Clinical Dosimetry

Recommendations of the International Commission on Radiological Units and Measurements

Handbook 87

United States Department of Commerce
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*In preparation.

UNITED STATES DEPARTMENT OF COMMERCE • Luther H. Hodges, *Secretary*

NATIONAL BUREAU OF STANDARDS • A. V. Astin, *Director*

Clinical Dosimetry

Recommendations of the International Commission on Radiological Units and Measurements (1962) ICRU Report 10d

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(This publication supersedes parts of H78. Handbooks 84 through 89 will completely replace H78. For an explanation, see the Foreword. Also, for a list of these titles, see page 3 of cover.)

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Foreword

The reports of The International Commission on Radiological Units and Measurements for a number of years have been published by the National Bureau of Standards in the Handbook series. In the past, each of the triennial reports of the ICRU represented a complete restatement of the recommendations of the Commission. Because of the increasing scope of its activities, however, the Commission in 1962 decided to modify the previous practice. It will issue a series of reports presenting the current recommendations of the Commission. Each report will cover a particular portion of the area of interest to the ICRU. This procedure will facilitate revision of ICRU recommendations and also spread out in time the workload of the Commission. This Handbook is one of the new series presenting the recommendations of the Commission on one aspect of the field with which the Commission is concerned. It presents recommendations agreed upon at the meeting of the Commission held in Montreux, Switzerland, in April 1962.

The National Bureau of Standards is pleased with its continuing opportunity of increasing the usefulness of these important reports by providing the publication outlet.

A. V. Astin, *Director.*

Contents

Preface

A. Scope

The International Commission on Radiological Units and Measurements (ICRU), since its inception in 1925, has had as its principal objective the development of internationally acceptable recommendations regarding:

(1) Quantities and units of radiation and radioactivity,

(2) Procedures suitable for the measurement and application of these quantities in clinical radiology and radiobiology,

(3) Physical data needed in the application of these procedures, the use of which tends to assure uniformity in reporting.

The Commission also considers and makes recommendations on radiation quantities, units and measurements in the field of radiation protection. In this connection, its work is carried out in close cooperation with the International Commission on Radiological Protection.

B. Policy

The ICRU endeavors to collect and evaluate the latest data and information pertinent to the problems of radiation measurement and dosimetry and to recommend the most acceptable values for current use.

Recognizing the confusion that exists in the evaluation of different radiological equipment and materials, the ICRU is studying standard methods of determination of characteristic data for the equipment and materials used in diagnostic and therapeutic radiology. This activity is confined to methods of measurement and does not include the standardization of radiological equipment or parts thereof.

The Commission's recommendations are kept under continual review in order to keep abreast of the rapidly expanding uses of radiation.

The ICRU feels it is the responsibility of national organizations to introduce their own detailed technical procedures for the development and maintenance of standards. However, it urges that all countries adhere as closely as possible to the internationally recommended basic concepts of radiation quantities and units.

The Commission feels its responsibility lies in developing a system of quantities and units having the widest possible range of applicability. Situations may arise from time to time when an expedient solution of a current problem may seem advisable. Generally speaking, however, the Commission feels that action based on expediency is inadvisable from a long-term viewpoint; it endeavors to base its decisions on the long-range advantages to be expected.

In 1955 the Commission entered into an official relationship with the World Health Organization (WHO). In this relationship, the ICRU will be looked to for primary guidance in matters of radiation units and measurements, and in turn WHO will undertake the worldwide dissemination of the Commission's recommendations. In 1960 the ICRU entered into consultative status with the International Atomic Energy Agency (IAEA).

The above relations with other international bodies do not affect the basic affiliation of the Commission with the International Society of Radiology.

The ICRU invites and welcomes constructive comments and suggestions regarding its recommendations and reports. These may be transmitted to the Chairman.

C. Current Program

A 2-week meeting of the ICRU was held in Montreux, Switzerland, April 2d to April 14, 1962. This meeting included the Main Commission and all of the Committees that had reports prepared for final approval. Some 70 persons attended. An additional meeting of the Commission and Committee Officers was held in Ottawa from August 21 to August 23, 1962, for the principal purposes of the preparation of the status report for the Xth International Congress of Radiology and the outlining of program objectives for the next several years.

Several meetings of committees or committee task groups have been held during the past 3 years. There were meetings of various task groups of the Committee on Standards and Measurement of Radiological Exposure—Paris in January 1961 and London in April and September 1961. The Committee on Radiobiological Dosimetry also held a meeting in April 1961. The ICRU was also represented at a meeting of the Consultative Committee on Ionizing Radiation of the International Committee of Weights and Measures at Sèvres in October 1961.

As noted in the last report, two joint committees had been established between the ICRU and the ICRP. The Joint Committee on RBE has met twice with ICRU participation. The Committee on Methods and Instruments for Radiation Protection has not met.

Upon the request from the United Nations Scientific Committee on the Effects of Atomic Radiations, the ICRU and the ICRP agreed to undertake a second study dealing with the Medical and Physical Parameters in Clinical Dosimetry. This committee met in New York for one week in September 1959 and for a week in Stockholm in June 1960. A report of this study entitled "Exposure of Man to Ionizing Radiation Arising from Medical Procedures with Special Reference to Radiation Induced Diseases, An Inquiry into Methods of Evaluation," was published in Physics in Medicine and Biology, **6,** No. 2, 199 (Taylor & Francis, Ltd., London, England, Oct. 1961).

Reports and recommendations of the ICRU, originally designed for medical applications, have come into common use in other fields of science, particularly where "dosimetric" considerations are involved. For this reason the committees have included in their membership some scientists having competence outside of the medical radiology field. Material in the report is designed to meet physical, biological, and medical requirements wherever possible.

This has introduced a small problem in terminology. The name of the Commission includes the term "radiological". In many European countries the term "radiological" is taken as inclusive of both the physical and biological sciences. In other countries, the United States for example, "radiological" appears to carry the primary connotation of relationship to medicine. It therefore may be desirable to change the name of the Commission from "Radiological" to "Radiation." It is believed that this would be properly understood by all concerned. The question has been debated by the Commission, but final action is being delayed for future consideration.

D. The Current Series of Reports

Hitherto, the triennial reports of the ICRU have been published in single volumes. However, the reports are now becoming too extensive, and in some cases too specialized, to make a single publication practicable. Beginning with this 1962 series, the ICRU reports will be issued in smaller entities, each dealing with a limited range of topics. The 1962 series supersedes the 1959 report. Revisions of the 1962 series will be undertaken individually as circumstances warrant. A full listing of ICRU recommendations, including the present series, is given on page iii of the cover of this report.

The current report series include revision of much of the material that appeared in the 1959 report in addition to a number of new topics. The following summary indicates some of the highlights of the current report series.

Radiation Quantities and Units (Report 10a)—One of the most important changes is the revision of the section on quantities and units. This revision resulted from the thorough study by an Ad Hoc Committee on Quantities and Units. It includes new names for certain quantities and clarified definitions for others. It presents a system of concepts and a set of definitions which is internally consistent and yet of sufficient generality to cover present requirements and such future requirements as can be foreseen.

Physical Aspects of Irradiation (Report 10b)—This report deals broadly with the physical aspects of irradiation with a considerable amount of new material added since the 1959 report. It includes an extensive discussion of the various techniques for the measurement of absorbed dose as well as exposure. Characteristics of radiation instrumentation are covered in some detail including the more sophisticated work on standards. The section on spectra has been up-dated and a new section added on neutron measurements and standards. Available data for stopping power ratios and the average energy (W) required to produce an ion pair in a gas have been reviewed. On the basis of this review it has been necessary to modify the previous ICRU tables for these factors. This modification amounts to about 1 or 2 percent change in stopping power ratios and up to 1 percent in W.

Radioactivity (Report 10c)—The portions of the report dealing with direct and relative measurements of radioactivity and the availability and requirements for radioactivity standards, and the parts dealing with the techniques and measurements of radioactivity in hospitals and biological laboratories are revisions of the 1959 report, embracing a review of the developments that have occurred since that report and bringing up to date the material included. In addition, a new section on low level radioactivity in materials as related to the problems of radiological measurements has been added. This topic is important because of the problems arising from the contamination, or possible contamination, in the last decade of a great many of the materials used in the construction of counting equipment, shields, and in the reagent chemicals employed in radioactivity measurements.

Radiobiological Dosimetry (Report 10e)—This report deals primarily with radiobiological dosimetry, and considers methods of improving the accuracy and intercomparability of absorbed dose measurements in radiobiology. It is in effect a handbook for the experimental radiobiologist. It emphasizes the great importance of planning the experimental work in a way which makes the dosimetry easier and more accurate and it illustrates how this can be done.

Methods of Evaluating Radiological Equipment and Materials (Report 10f)—This is the first of a new group of ICRU reports dealing with methods of evaluating radiological equipment and materials. It includes a revised discussion on the measurement of focal spots and new sections on grids, image intensifiers, and body section equipment.

E. Operating Funds

Throughout most of its existence, the ICRU has operated essentially on a voluntary basis, with the travel and operating cost being borne by the parent organizations of the participants. (Only token assistance was available from the ISR.) Recognizing the impracticality of continuing this mode of operation on an indefinite basis, operating funds were sought from various sources in addition to those supplied by the International Society of Radiology.

Prior to 1959, the principal financial assistance to the ICRU had been provided by the Rockefeller Foundation which supplied some $11,000 to make possible various meetings. In 1959 the International Society of Radiology increased its contribution to the Commission to $3,000 to cover the period until the Xth Congress. In 1960 the Rockefeller Foundation supplied an additional sum of some $4,000 making possible a meeting of the Quantity and Units Committee in 1960.

In 1960 and 1961 the World Health Organization contributed the sum of $3,000 each year to the Commission for carrying forward its work. This was increased to $4,000 in 1962. It is expected that this sum will be allocated annually, at least for the next several years. In addition, the WHO has provided substantial assistance to the Commission in providing meeting space, secretarial services, etc., for the meetings held in Geneva and Montreux.

In connection with the Commission's Joint Study with the ICRP, the United Nations allocated the sum of $10,000 for the joint use of the two Commissions for the purpose of carrying out their second study. This fund has been administered by the ICRP.

The most substantial contribution to the work of the ICRU has come from the Ford Foundation through the particular efforts of Dr. Paul Pearson. Effective in December 1960, the Ford Foundation made available to the Commission the sum of $37,000 per year for a period of 5 years. This money is to be used for such things as travel expenses to meetings, for secretarial services, and other operating expenses. To a large extent, it is because of this grant that the Commission has been able to hold the several meetings considered to be necessary to move forward actively with its program.

The International Atomic Energy Agency has allocated the sum of $6,000 per year for use by the ICRU. It is expected that this sum will be allocated annually at least for the next several years.

A valuable indirect contribution has been made by the U.S. National Bureau of Standards where the Secretariat has resided. The Bureau has provided substantial secretarial services, reproduction services and traveling costs in the amount of several thousands of dollars.

The Commission wishes to express its deep appreciation to all of these and other organizations that have contributed so importantly to its work.

F. Composition of the ICRU

(a) It is of interest to note that the membership of the Commission and its committees for the period 1959–62 totals 139 persons drawn from 18 countries. This gives some indication of the extent to which the ICRU has achieved international breadth of membership within its basic selection requirement of high technical competence of individual members.

(b) The membership of the Main Commission during the preparation of this report was as follows:

Lauriston S. Taylor, Chairman	United States
L. H. Gray, Vice-chairman	United Kingdom
H. O. Wyckoff, Secretary	United States
K. K. Aglintsev	U.S.S.R.
A. Allisy	France
R. H. Chamberlain	United States
F. Ellis	United Kingdom
H. Fränz	Federal Republic of Germany
H. E. Johns	Canada
W. J. Oosterkamp	Netherlands
B. Rajewsky	Federal Republic of Germany
H. H. Rossi	United States
M. Tubiana	France

G. Composition of Committee Preparing Initial Draft of Present Report

J. W. Boag, Chairman, ICRU Committee III, "Measurement of Absorbed Dose and Clinical Dosimetry".
M. Tubiana, Chairman, Committee III–B, "Clinical Dosimetry"
R. H. Chamberlain
F. Ellis
H. E. Johns
W. J. Meredith
R. Robbins

Consultants

M. Lindgren
M. D. Schulz

H. The Present Report

Much of the present report is addressed principally to the clinician engaged in radiotherapy although it is hoped that it may also serve as a useful guide to physicists entering the field. The report attempts to explain the principles underlying good radiotherapeutic techniques in words rather than in formula, and to make recommendations based on these principles.

It is realized that the present survey is far from complete. Thus, practically no mention is made of superficial X-ray therapy, of interstitial and intracavitary techniques, of therapy with isotopes (except in appendix IV), or of the use of electron beams. There are doubtless other important matters which have been overlooked, and perhaps some which have been treated at too great length. Comments from the readers would assist the Commission when the time comes to issue a revised edition of the Report.

The Reports of the ICRU have in the past been concerned chiefly with the definition and measurement of the various physical quantities which are needed in any quantitative use of radiation in medicine or biology. These quantities have evolved considerably in the last two decades, and the variety and precision of the methods of measuring them have also increased. During the last three years, a special sub-committee set

up by the Commission has been studying the more practical questions which arise in the specification, delivery, and recording of radiation dose in clinical work. The present report summarizes the findings and recommendations of that Committee.

One of its first tasks was to undertake a critical survey of published data on dose distributions, methods of calculation, and the like. To facilitate this and to draw upon critical and constructive assistance from as wide a circle of experts as possible, a special study group was set up, under the joint auspices of the WHO, the IAEA, and the ICRU. This group met in Geneva in April 1961 and prepared papers which have been used as the basis for much of the present report. In addition to the members of Committee III–B of the ICRU, the following persons took part in that meeting, the expenses of which were shared equally between the three sponsoring organizations:

C. B. Braestrup
M. Cohen
M. J. Day
R. L. Dobson
A. Dutreix
D. E. A. Jones
J. S. Laughlin
B. Markus
V. A. Petrov
K. C. Tsien

Clinical Dosimetry

International Commission on Radiological Units andMeasurements (ICRU) Report 10d 1962

I. Introduction

The effects of ionizing radiations on a patient are largely determined by the magnitude of the absorbed dose and its distribution in both space and time. They are, undoubtedly, also considerably influenced by such factors as the chemical environment of the irradiated cells (oxygen tension, pH etc.), and by the microscopical distribution of the absorbed energy (LET), which depends on the nature and quality of the radiation. The systematic study of these latter effects is, at present, mainly the concern of the radiobiologist, and no attempt can be made to take account of them in this report which is largely concerned with the magnitude and the pattern of the absorbed dose within the patient. For good radiotherapy it is essential to have adequate knowledge of these parameters, and methods of varying complexity have been used to estimate them. Generally speaking, the more complex the methods the greater the accuracy achieved. For palliative treatments simple calculations may suffice; for radical treatments much greater accuracy is desirable.

The simplest approach is to regard the patient as equivalent to a cube of unit density material, and to determine only doses along the central rays of the treatment beams. For this, some method of determining the radiation output has to be used, along with tables of central percentage depth dose values. In this way doses at only a few points in the treatment scheme can be determined. More information may be obtained by the use of isodose charts instead of merely central ray data.

It must be recognized, however, that the patient is neither cubic nor homogeneous and that the doses arrived at on such assumptions are only approximations to the dose within the patient. Corrections may, however, be applied for extra absorption in bone, or extra transmission through the lung, for the curvature of the surfaces through which beams enter the patient, or for the fact that the part of the body being treated is much smaller than the phantom in which the basic data was measured.

Unfortunately, there are considerable differences between different centers in the methods used for estimating dose, and often a striking lack of agreement between the basic data employed. As a result two departments using identical treatment techniques and radiation, may report quite different doses for the same effect, simply because of their different methods of dose computation. Further confusion is often added by attaching different meanings to commonly used technical terms. This report endeavours to bring some uniformity into the methods of assessing dose by suggesting standard procedures and sources of data, the use of which will not only increase the accuracy of any dose statement, but will increase comparability of work in different centers

Happily, there is reasonable agreement on the general concepts of dose quantities and units and these are outlined in our first section. This is followed by some definitions of frequently used terms. Then a technique for output calibration is described which eliminates much of the difference between published values of central percentage depth doses. Depth dose data and isodose curves which can be used with confidence are listed and methods of obtaining this type of information by measurement or calculation, should it not be available in published form, are discussed. Considerable attention is given to methods by which allowances can be made for differences between the body and a homogeneous phantom, and the confidence that may be placed in both the basic data and the corrections is indicated.

All too often the value of published articles is largely lost because of failure to provide enough information to allow others to reproduce the methods discussed, or to evaluate them critically. Proposals are, therefore, put forward about the detailed information necessary when radiation treatments are recorded or reported. The most common errors in clinical dosimetry are listed and brief comments and recommendations are made.

II. Fundamental Concepts, Quantities and Units

The physical properties of a beam of x or gamma rays are often summarized in the two concepts *intensity* and *quality*. The intensity (which is defined precisely in the ICRU Report 10a (1962) may be loosely described here as the rate of flow of X-ray energy along the radiation beam per unit of its cross sectional area. This concept is of importance in theoretical physics, but of less use in medical radiology, where interest is concentrated on the interaction between the radiation beam and the patient, that is to say, on the energy deposited in tissues by the secondary electrons liberated as a result of the irradiation. To describe this interaction the concept of *absorbed dose* has been adopted. This concept is discussed and defined in the ICRU Report 10a, and precise definitions are given in Appendix I of the present report, but in the present consideration of clinical applications the following simplified definition will suffice.

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Absorbed dose of any ionizing radiation is the energy imparted to matter by ionizing particles per unit mass of irradiated material at the place of interest. The unit of absorbed dose is the rad.

$$1 \text{ rad} = 100 \text{ erg}/g = 1/100 \text{ Joule}/kg.$$

Methods of measuring absorbed dose, together with the data needed to evaluate these measurements, are given elsewhere. (ICRU Report 10b, 1962.) In this report, we shall assume that appropriate measuring instruments are available and shall discuss only the corrections or precautions which may be required when using these instruments in phantoms or in clinical work.

Although the concept of intensity has little application in the field of medical radiology, it is useful to have some way of expressing the radiation output of sources and advantageous if this can serve as a step in the determination of the absorbed dose received by a patient exposed to the beam. The concept of *exposure*, which expresses the ability of the beam to ionize air, serves both the desired ends. Its unit is the roentgen and is based on well established and convenient measuring techniques.

For x rays of energy below a few Mev (above these it is difficult with present techniques to measure exposure) a knowledge of exposure at any point in a material is sufficient to allow absorbed dose to be calculated by methods which will be briefly described here and which are set out more fully in ICRU Report 10b (1962).

The term "*quality*," as applied to ionizing radiation, attempts to summarize its power to penetrate through matter. This power depends on the energies of the photons which make up the beam and many methods have been used to describe it, including:

(1) The *electrical potential* through which the electrons striking the target have been accelerated. This is expressed in kilovolts or megavolts (in the case of equipment with varying voltages the peak voltage is stated) and the maximum energy of the photons corresponds to this accelerating potential.[1] In the case of accelerators employing multiple acceleration one can state, by analogy, the equivalent electrical potential through which the electrons would have to fall in order to attain the the energy they possess when they strike the target.

(2) *The half value layer* (*HVL*). This is the thickness, or the surface density, of a layer of a specified material which attenuates the beam to such an extent that the exposure rate is reduced to one half, under narrow beam conditions. For x-rays between 50 and 150 kv the HVL is usually stated in mm of aluminium; between 200 and 400 kv copper is used, while for more penetrating radiations the HVL is usually stated in mm of lead.

(3) *HVL and homogeneity coefficient.* The ratio of the HVL and the additional thickness of the material needed to reduce the beam to one quarter of its original exposure rate, is called the homogeneity coefficient. It is unity for monoenergetic photons.

(4) *Spectral distribution.* This is the most complete description of the composition of the beam. A knowledge of it would be necessary for precise calculations of the quality and quantity of scattered radiation at various parts of a medium, which might be needed in high accuracy estimations of absorbed dose in tissues which are not water equivalent. This method of quality specification is not appropriate to clinical radiotherapy.

Recommended methods of quality specification. A distinction may here be made between gamma-ray beams from radioactive nuclides, with their relatively small number of different photon energies, and the wide spectrum from an x-ray source.

Gamma-ray beams. Often it is sufficient to *specify the type of nuclide* or nuclides and their relative amounts in the source. There may, however, be cases where the scattered photons, or the bremsstrahlung, from the source and its surroundings are not negligible, or where low energy gamma-ray beams can be modified by filtration. In such cases *the HVL of the emerging beam should also be stated.*

X-ray beams. The complex spectrum from an x-ray source is more difficult to specify and more than one parameter is often needed, for example, for x rays in the range up to 2 Mv where filtration may considerably modify the beam quality. Above this voltage changes of filtration alter the quality very little and a statement of the generating voltage will suffice. This is also true for situations where the radiation is highly attenuated, as in x-ray diagnosis and in radiation protection problems.

The following methods of quality specification for x-ray beams are therefore recommended for general clinical use:

(a) For radiotherapy up to 2 Mv. State the kv or Mv and HVL.

(b) For radiations above 2 Mv and in the diagnostic range. State the kv or Mv only.

[1] There has been hitherto a lack of uniformity in the nomenclature used for describing megavolt x radiation. For monochromatic radiation it is, of course, correct and unambiguous to refer to the quantum energy of the radiation. Thus Co^{60} emits two gamma rays of energies 1.17 Mev and 1.33 Mev respectively, and the primary beam from a cobalt source contains quanta of these two types in fixed proportions. The x radiation produced when electrons hit the target, however, has a continuous spectrum of energies (Bremsstrahlung) in addition to a few monochromatic components (characteristic radiation). When electrons which have been accelerated through 1.3 Mev strike a thick target, very few of the resulting x-ray quanta have energies close to 1.3 Mev, and the radiation is much less penetrating than that from a Co^{60} source. The *average* quantum energy as determined by the penetrating power of the radiation is between 40 and 45 per cent of the maximum possible energy for the filtration normally employed clinically. It is therefore misleading to describe radiation as "1.3 Mev X-radiation." The alternative proposal, which is embodied in the definition given above, is to quote the accelerating potential applied to the x-ray tube, as has long been the practice at voltages up to about 2 Mv. The principal objection to extending this practice to betatrons, linear accelerators and the like is that in such machines the full potential does not appear anywhere, since the electrons gain energy by repeated transits through a smaller potential, or by being held in a constant electric field. This is, however, a matter of the engineering design of the apparatus. In such machines the x-ray beam is identical with that which would be obtained in an apparatus in which the electrons had fallen once through the full potential.

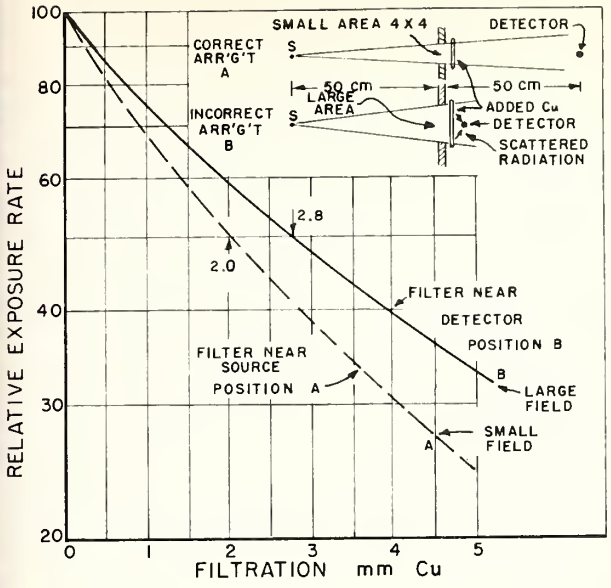

FIGURE II.1. *Geometry for and results from attenuation measurements.*

Arrangement A shows correct method for HVL determination which gives somewhat lower values than the incorrect Arrangement B.

The measurement of HVL. (See also, ICRU Report 10b, 1962.) Farr [1] and Johns [2] have demonstrated that to obtain a correct measurement of the HVL, a narrow beam and a large distance between absorber and detector must be used. Figure II.1 (arrangement A) shows a simple and practical arrangement, which can easily be reproduced in any department and satisfies Farr's conditions. A small treatment cone giving a 4x4 cm field at 50 cm is used. Layers of the material in which the HVL is being measured are added at the position indicated and a small detector is placed 50 cm farther from the source. Under these conditions, a minute fraction of the scattered photons from the absorber are received by the detector and the condition can be called "narrow beam". The graph shows that under these conditions in the example given a layer of copper 2.0 mm thick reduces the exposure to 50 percent so the HVL is 2.0 mm Cu.

On the other hand, if arrangement B is used for the experiment, an appreciable proportion of the photons scattered from the absorber can enter the detector, so that the exposure does not decrease as rapidly with added absorber, and curve B is obtained, from which an incorrect HVL of 2.8 mm Cu would be estimated. If a very precise value of HVL is required, the method described by Trout, Kelley, and Lucas [3] should be used, though the method just described is adequate for all clinical purposes, especially as percentage depth doses vary slowly with radiation quality. A fuller account of the problems of quality specification and of half value layer measurement will be found in ICRU Report 10b (1962).

References

[1] R. F. Farr, The specification of roentgen ray output and quality, Acta Radiol. **43,** 152 (1955).

[2] H. E. Johns, The Physics of Radiology (Charles C. Thomas, Springfield, Ill., 1961).

[3] E. D. Trout, J. P. Kelley, and A. C. Lucas, Determination of half value layers, Am. J. Roentgenol. **84,** 729 (1960).

III. Definition of Terms

The principal object of defining carefully the meanings of terms which are commonly used to describe the apparatus or techniques of radiotherapy, is to permit experience gained in one center to be readily transmitted to and applied in other centers.

A. Accessory Equipment Used to Modify the Characteristics of the Beam

1. *Filter.* A filter is an absorbing material inserted into the beam in order to modify its spectral composition, to suppress particular components of a mixed beam, (for example, a beam containing both beta and gamma rays), or to modify the spatial distribution of intensity. The filtration of the x-ray beam due to its passage through the tube wall, the window of the housing, etc., and any other materials permanently situated between the target and the collimator, is called the inherent filtration. It is often expressed as the equivalent thickness of a given substance which, if inserted as a filter immediately in front of the target, would produce a radiation beam of the same quality and intensity as that which emerges from the apparatus.

2.1 *Beam flattening filter.* A filter designed to make the intensity reasonably constant across the beam.

2.2 *Wedge filter.* A filter of graduated thickness which causes a progressive decrease in the intensity across the whole or part of the beam.

2.3 *Compensating filter.* A filter designed to modify the dose distribution within the patient in any desired manner, i.e., this filter may take account of the size, shape, orientation or position of the patient.

3. *Scattering foil.* A foil introduced into an electron beam to broaden the beam and thus render the dose distribution more uniform over a plane normal to the beam axis.

4. *Collimator.* A diaphram or system of diaphragms made of an absorbing material which are designed to define the dimensions and direction of a beam of radiation.

5. *Applicator (or treatment cone).* A structure which indicates the extent of the radiation field at a given distance from the source, and which may or may not itself incorporate the collimating diaphragms. Applicators may be either closed-ended or open-ended.

B. Geometrical Characteristics of the Radiation Beam

1. *Central ray.* The straight line passing through the center of the source and the center of the final

beam limiting diaphragm. (The latter means the center of symmetry of the plane figure formed by that edge of the diaphragm system which defines the beam. If the final diaphragm has no center of symmetry, a more detailed description is necessary.)

2. *Principal plane.* A plane which contains the central ray and, in the case of rectangular section beams, is parallel to one side of the rectangle.

3. *Geometrical edges of the beam.* The lines joining the center of the anterior face of the source to the diaphragm edges furthest from the source.

4. *Geometrical field size.* The geometrical projection on a plane perpendicular to the central ray, of the distal end of the limiting diaphragm as seen from the center of the front surface of the source. The field is thus the same shape as the aperture of the collimator.

The geometrical field size can be defined at any distance from the source. Positions of special interest are at the skin surface, at a distance corresponding to the center of the target volume, or at the axis of rotation in moving beam therapy.

The geometrical field as defined here will be similar in size and shape but not identical with the "physical field" which some workers have defined as being outlined by the 50 percent isodose curve at the depth of the maximum. It will be larger than the fields defined in terms of the 80 or 90 percent isodose curves, and those who have been accustomed to these conventions should take special note of this, as should those who have previously been accustomed to x rays below 400 kv where the surface dose at the field edge is seldom below 85 percent of the central dose.

The geometrical field size is not the same as the *port of entry* of beam into the patient. Since the contour of the patient may be irregular, the entrance port, too, may have a very irregular shape, and it cannot readily be used to define the geometrical properties of the beam.

For electron beams geometrical field size is not an adequate parameter. Where the beam is defined by a collimator which is in contact with the surface, it is recommended that the inside dimensions of the collimator be taken as the field size. Where the defining system is not in contact with the surface, the field size is defined by the position of the 90 percent dose contour, (relative to the maximum dose) at a depth of 1 cm below the skin surface.

5. *Geometrical penumbra.* The geometrical penumbra is the region of free space which is irradiated by primary photons coming from only part of the source.

By analogy, the transmission penumbra is the region of space irradiated by photons which have traversed only part of the thickness of the collimator, i.e., the part of the collimator at its lower edge.

6. *Penumbra width.* The width of the penumbra in a plane perpendicular to the central ray at any distance of interest from the source.

C. Characteristics of the Irradiation Procedure

Fixed fields

1. *Source-surface distance (SSD).* The distance measured along the central ray from the front surface of the source to the surface of the irradiated object. (N.B. The above definition can apply to x-ray as well as to gamma-ray sources.) The use of the older term "focus surface distance" (FSD) is to be discouraged.

Two other dimensions are frequently of interest in describing the properties of a beam. These are (1) the source-collimator distance, measured along the central ray from the front surface of the source to the distal end of the collimating diaphragm, and (2) the diaphragm to surface distance measured along the central ray from the distal end of the collimating diaphragm to the surface of the irradiated object.

2. *Angle of incidence.* The angle between the central ray of the incident beam and the normal to the irradiated surface is called the angle of incidence.

3. *Opposed beams.* Two beams which enter the patient in opposite directions and whose central rays are coincident.

Moving beam therapy. Treatment by one or more radiation beams, which move in relation to the patient during irradiation.

When the source is moved it often describes a circle or a circular arc, and the plane containing this circle is then called the plane of rotation. Some specialized terms used in moving beam therapy are discussed in section VII.

D. The Distribution of Dose

1. *Dose distribution.* A representation of the variation of dose with position in any region of an irradiated object. The dose distribution may be measured using detectors small enough to avoid disturbing the distribution, or in simple cases, it may be calculated and expressed in mathematical form.

2. *Tissue-equivalent material.* A liquid or solid whose absorbing and scattering properties for a given radiation simulate as closely as possible those of a given biological material, such as fat, bone or muscle. For muscle or soft-tissue, water is usually the best tissue-equivalent material.

3. *Bolus.* Additional tissue-equivalent material placed round the irradiated object to provide extra scattering or absorption.

4. *Phantom.* A volume of tissue-equivalent material either large enough to provide adequate scatter or constructed to resemble some special object, such as part of the human body, for the purpose of measuring a dose distribution.

A phantom made of one material only is called a homogeneous phantom. When a phantom simulates the heterogeneity of the human body, it is called a heterogeneous phantom.

5. *Target volume.* The region in the patient to which it is desired to deliver a specified dose for

treatment purposes. (This is usually the tumor or lesion with its associated and inherent normal structures.)

6. *Surface absorbed dose.* The absorbed dose delivered by a radiation beam at the point where the central ray passes through the superficial layer of the phantom or patient. (N.B. In an asymmetrical beam of radiation the surface absorbed dose thus defined may not be the maximum absorbed dose at the surface.)

7. *Peak absorbed dose.* The maximum value of the absorbed dose which occurs along the central ray.

The peak absorbed dose is situated at a depth of about 0.2 cm for Cs^{137}, 0.5 cm for Co^{60}, 1 cm for 4 Mv x radiation, and 4 cm for 25 Mv x radiation. The exact position of the peak depends on the type of collimator, and on whether a secondary electron filter is used. It also depends on the SSD and the field size. (N.B. In an asymmetrical beam such as that produced by a wedge filter, the peak absorbed dose may be less than the maximum absorbed dose, which occurs off the central ray.)

8. *Given (or applied) dose.* The surface dose (for radiation below 400 kv) or the peak dose (for harder qualities) delivered by one beam in a complete treatment, or a treatment session.

9. *Build-up.* In a material irradiated by a beam of x rays or gamma rays, the increase in absorbed dose with depth below the surface is called the build-up. This is due to an increasing production of secondary electrons in the material, as well as a build-up of scattered photons due to multiple scattering in broad beams of radiation. For high energy beams the former process is the more important.

10. *Build-up factor.* In a beam of high energy x or gamma rays the ratio of peak absorbed dose to surface absorbed dose is called the build-up factor.

11. *Output (normalised exposure rate).* Output is a term used to describe exposure rate, at a specified point under standardised conditions. If the point of measurement is in free air, the output is expressed in roentgens per minute, or similar units of convenient magnitude. If the point is at the surface of, or at the peak dose depth in a phantom, the output is expressed in roentgens per minute (below a few Mv) or rads in tissue per minute.

The output measurement is liable to be influenced by scatter from the collimator and from objects near the measuring instrument, and the "standard conditions" must, therefore, be specified very carefully.

12. *Scatter factor, including back scatter factor.* The ratio of the exposure at a reference point in the phantom to the exposure at the same point in space under similar conditions of irradiation in the absence of the phantom. For up to 400 kv radiation the reference point is to be taken at the intersection of the central ray with the surface, and the scatter factor is usually called *"back scatter factor"*, while for radiation above 400 kv the reference point is to be taken at the position of the peak dose. In measuring the scatter factor the phantom should have a cross section of 30x30 cm and extend at least 10 cm beyond the depth at which the factor is being measured.

13. *Percentage depth dose.* The percentage depth dose in an irradiated body is the ratio (expressed as a percentage) of the absorbed dose, D_x, at any depth x to the absorbed dose, D_0, at a fixed reference point on the central ray.

$$\text{percent depth dose} = 100 \times \frac{D_x}{D_0}$$

For x radiation produced at up to 400 kv the reference point is at the surface. For x radiation above 400 kv and gamma teletherapy, the reference point is at the position of the peak absorbed dose. For rotation therapy it is often convenient to take the center of rotation as the reference point.

14. *Isodose surface.* A surface on which the absorbed dose is constant.

15. *Isodose curve (or contour).* A line along which the absorbed dose is constant. (N.B. For x rays up to about 400 kv the isodose surfaces or curves may alternatively be drawn as surfaces or curves of constant exposure.)

16. *Isodose chart.* A set of isodose curves, usually drawn for regular intervals of absorbed dose or of percentage depth dose, which represent the distribution of dose over a particular surface within the irradiated body.

17. *Wedge isodose angle.* The complement of the angle which the isodose curve for 50 percent of the peak absorbed dose in a wedge isodose chart, makes with the central ray,[2] in a principal plane of the field.

18. *Tissue air ratio (TAR).* The ratio of the absorbed dose at a given point in a phantom to the absorbed dose which would be measured at the same point in free air within a volume of the phantom material just large enough to provide the maximum electronic build-up at the point of measurement.

19. *Exit dose.* The absorbed dose delivered by a beam of radiation at the surface through which the beam emerges from the phantom or patient.

20. *Transit dose.* A measure of the primary radiation transmitted through the patient and measured at a point on the central ray at some point beyond the patient.

21. *Integral dose.* The total energy absorbed in a specified region. (It is the integral with respect to mass of the absorbed dose throughout the region.) Integral dose may be measured in gram-rads or in any convenient units of energy. The gram-rad itself is equal to 100 ergs.

22. *Integral dose efficiency factor.* The ratio of the integral dose in the target volume to the total integral dose to the patient.

[2] The reference point for the isodose curves should be as recommended in [13] of this section.

23. *Session.* A treatment or group of treatments in one visit.

24. *Fractionation.* The splitting of a dose into a number of short sessions given over a longer period than would be required if the dose were given continuously in one session at the same dose rate.

25. *Fraction.* A single session in a fractionated treatment.

26. *Overall time.* The total time elapsing from the beginning to the end of a session or of a series of sessions if the treatment is fractionated.

IV. Practical Calibration Procedures and Determination of Central Ray Doses

A. Introduction

The problem of the computation of dose at some point or points in an actual treatment is the first to be considered. For this it is necessary to know the output of the apparatus at some point, and to know the distribution of the radiation within the patient, or more frequently within a homogeneous phantom representing the patient. This latter information is usually presented as central percentage depth dose tables or isodose curves.

The technique of output determination will be discussed fully below. It is essential to use a suitable dosimeter, which may be chosen from one of the many commercially available types. The chosen instrument should preferably have a small ionization chamber (a diameter of 0.5 cm and a length of up to 1.5 cm can be regarded as very satisfactory), though this is not essential if it is only to be used for the "in-air" exposure method of calibration mentioned below. The chamber sensitivity should be as independent as possible of radiation quality over a wide range, and the instrument should require as little correction as possible for effects like the so-called "stem effect" (Braestrup and Mooney [1],[3]) the magnitude of which may vary with radiation quality and also with the previous irradiation history of the instrument. (See figure IV.1 for the set-up used to determine the stem leakage.) It is also most important that the instrument should be calibrated either at a national standardizing laboratory, or against another instrument that has been so calibrated, for all radiation qualities for which it will be used. Moreover, such calibration should be repeated at least every two years, and in the intervening period frequent sensitivity checks should be made with the aid of a suitable radioactive source. Only in this way can constancy of dose calibration be assured.

[3] When the stem of a condenser dosimeter is irradiated with high energy radiations, the indicated exposure rate may be higher than it should be because of the production of a leakage current through the insulator and ionization in extra-cavitary air gaps. A simple test may be applied which will indicate whether a particular instrument shows this effect, and will give an indication of its magnitude. Using a field, just wide enough to include the ionization chamber without its being in any penumbra, and about twice as long as the chamber, measure the output first with the chamber axis along the field, A in figure IV.1, and then with the chamber axis at right angles to the long axis of the field, B as shown in figure IV.1. Any difference in the two readings will be due to the stem effect. The variation of the effect with the extent of the stem irradiation can be found by repeating the experiment with fields of different length.

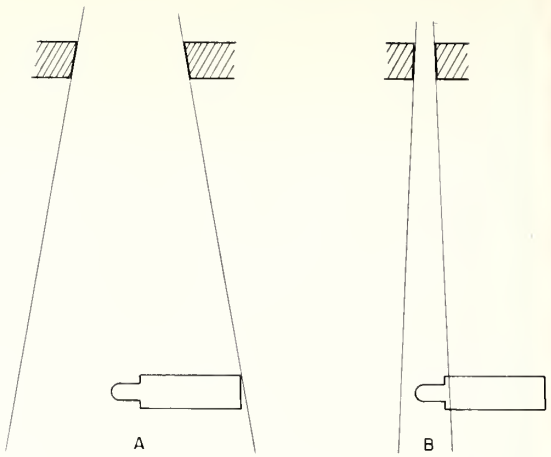

FIGURE IV.1. *Arrangement for testing the stem leakage of condenser dosimeters.*

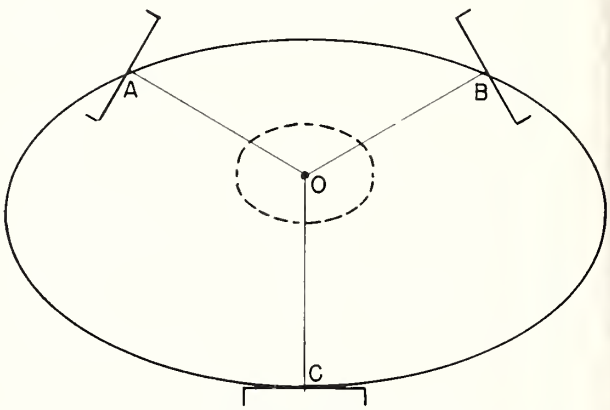

FIGURE IV.2. *An example of a three-field treatment.*

The target volume is within the dotted line; the central ray of each of the fields is indicated by lines AO, BO, and CO.

To illustrate the recommended methods of dose computation, it is best to consider an example and for this a three-field treatment has been chosen. In this, it is assumed that the target volume and its relationships to the three treatment fields are known.

The minimum information on dose that can be contemplated is the dose delivered to the center of the target volume, which is usually (as in the example shown in fig. IV.2), the point of intersection of the central rays of the treatment beams. To calculate this, central percentage depth dose values are required. Normally, however, information about dose in other parts of the target volume, or at important points outside it, such as at the skin, will also be required, and for this isodose curves have to be used. The present section will deal only with the way in which central ray information is applied to a homogeneous "patient". Later chapters will deal with the uses of isodose curves in fixed and moving beam therapy, and also the correction which should be made to allow for the actual size, shape and inhomogeneities of the patient.

B. Absorbed Dose Inside a Phantom

To find the absorbed dose-rate at any point (such as the point 0 in fig. IV.2) in a phantom, we must know, among other things, the output at some specified point, for the equipment, and the percentage depth dose at the point in question. Two main approaches are at present in use and may, respectively, be described as the "in air" exposure method and the "surface dose with full back scatter" method. Although the basic principles involved are the same for all radiation energies, attention here will first be focused on x rays up to about 400 kv. Any special considerations applying to higher energy radiations will be dealt with later.

In the first of these two methods, the exposure rate in air is determined for the center of the field and at the working SSD either by measurement made directly at that point (where open-ended applicators or diaphragm-limited fields are being used) or by computation, based on measurements made at some distance from the point of interest (where closed-ended applicators are being used). A fuller discussion of this matter may be found in ICRU Report 10b (1962).

For the second method, the dosimeter is laid in a groove in the phantom surface, and with its center at the surface level. In this way, it can record both the primary and scattered radiation reaching the surface. Especially when closed-ended applicators are used, a small inverse square law correction has to be applied because the applicator is inevitably separated, by half the thickness of the measuring device, from the surface with which it should be in contact.

The output established by one or other of these methods is then multiplied by (a) the percentage depth dose for the depth of the point in question and for the radiation parameters (SSD, Quality and Field Size) in use; by (b) the appropriate factor for the conversion of exposure into absorbed dose, and in case of output measured by air, by (c) the appropriate backscatter factor.

Both these methods can yield accurate estimates of the absorbed dose-rate in tissue, *provided the method of output determination and the beam defining system are the same as those used for the production of the percentage depth dose values.* Even in one department it is impossible to apply to one x-ray source, depth dose data obtained on a different type of source, without risk of considerable inaccuracy.

Now many departments have neither the staff nor the facilities to undertake the extensive series of measurements entailed in the production of all the percentage depth dose data needed. Therefore, they use published material often without having much information of the methods of output determination and beam collimation used in its production. In this way differences of 10-percent in quoted doses can exist between departments actually using identical treatments. [2]

The problem is further complicated, however, because although many tables of percentage depth doses have been published for radiation qualities and SSD in general use, the most striking feature of the data for radiation below 400 kv is its lack of agreement. Figure IV.3 shows a number of recent sets of percentage depth dose values for the same field size, HVL and SSD. Their disagreement is obvious and it is clearly desirable to reduce such uncertainties.

It is hardly practicable to lay down standard depth dose data for universal use, since there is no complete set of data from a single measuring source, and it would be impossible to determine objectively which set of values is the best. However, such a choice is unnecessary, for if the type of data shown in figure IV.3 is presented in another way, most of the difference disappears, and it is possible to bring all modern data into close conformity by a simple change of calibration technique.

It has long been recognized that measurements at an interface between two media are very difficult to make, and that surface or "in-air" measurements may be critically influenced by soft radiations from collimating systems or other material in the radiation beam. What has not been fully realized is, that because of these considerations, accurate dosage assessments can only be obtained with any given depth dose data provided the method of output measurement and the physical conditions of the beam (applicator type, filters, diaphragm positions, etc.) are the same as those used for the data measurements. If these are not repeated, errors of dose estimation may well result.

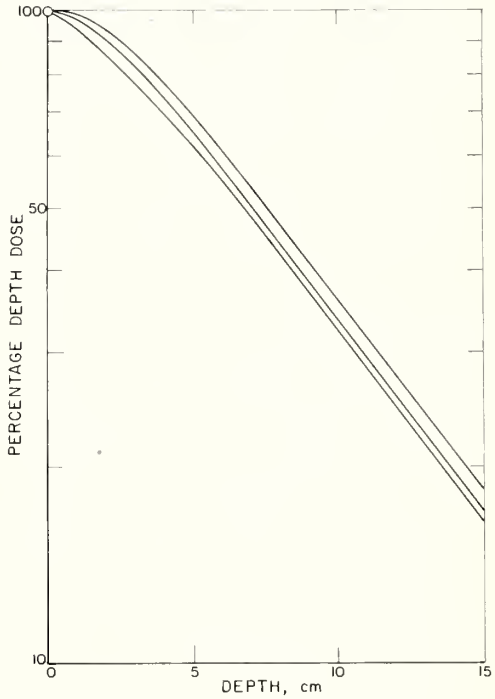

FIGURE IV.3. *Some published central percentage depth dose data.*

50 cm SSD, 2.0 mm Cu HVL normalized at surface.

On the other hand, measurements a few centimeters inside a phantom are easier to make, and are unlikely to be affected by collimator design, etc. In figure IV.4, the data presented in figure IV.3 are shown again, but whereas in figure IV.3 they are normalized at the surface; that is, they are presented in terms of surface measurements—in figure IV.4, they are normalized at 5 cm deep; that is, presented as if the basic comparison measurement was made at 5 cm deep. It will be seen immediately that all three sets of data are now in excellent agreement over the clinically important depths of 5 to 15 cm and, in fact, are now discordant only at or near the surface, which is relatively unimportant in well devised multi-field therapy.

Part of the disagreement shown in figure IV.3, it must be admitted, is due to the fact that one set of data is specifically for open-ended applicators, whereas the rest are for closed-ended applicators. This choice was deliberately made to demonstrate that, except near the surface, this difference has no real influence on the depth doses, and its effect is eliminated by taking a reference point below the surface, as in figure IV.4. The significance of these observations is that if we can measure the dose-rate at a point some way below the surface, the dose-rate computed for any other depth will be the same, whichever modern set of depth dose data is used. (Even older data obtained with instruments or in phantoms which would no longer be regarded as satisfactory can be used to give more accurate results than otherwise, if this approach is adopted.)

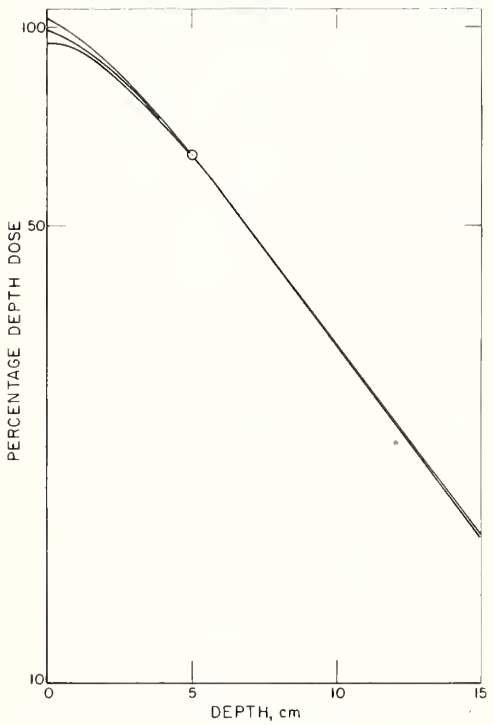

FIGURE IV.4. *The same percentage depth dose data as in figure IV.3 but normalized at 5 cm depth.*

It may be objected, however, that while the proposed method satisfactorily eliminates differences between relative doses everywhere except near the surface, it does not necessarily ensure that the absolute dose measurement will be satisfactory. Some widely used dosimeters were designed for in-air measurements and it has been shown [3] that the metal stem may shield the chamber from scattered radiation that should reach it when it is used in a phantom. Furthermore, the introduction of the chamber into the phantom displaces some material from around the point of measurement, thus changing the absorption and scattering there, while the unavoidable irradiation of part or all of the stem of the instrument may produce the stem effects already referred to. Errors so introduced do not all operate in the same direction and it is difficult to forecast their total effect. Studies were therefore undertaken to estimate the overall errors by measuring the ratio of the exposure dose at a fixed point in air to the exposure dose at a fixed point in a phantom [4] with a number of commercial ionization chambers (and some laboratory models), for a few field sizes and a number of radiation qualities. For radiations of HVL above 0.5 mm Cu the agreement, within one or two percent, was such as to make these instruments acceptable for the accurate measurement of dose-rate at 5 cm deep in a phantom.

We are thus led to proposing a new calibration technique which, nevertheless, can leave undisturbed the methods of output statement and depth dose calculation in use in any department.

C. Proposed Depth Dose Calibration Procedure

1. *For each applicator or field size,* a measurement of exposure rate is made at a depth of 5 cm in a suitable phantom, such as will be described later.

2. Any modern percentage depth dose tables, appropriate to the radiation conditions employed, may be used.

3. From the measurement and the percentage depth dose at 5 cm for the field size in question, the surface exposure rate "with full back scatter" can be calculated. From this and the back scatter factor, the "in-air" exposure rate can also be calculated. Either of these "output" values may then be used for the determination of treatment times or depth doses in the usual way.

An example may be helpful both to demonstrate the shortcomings of present methods and the advantages of that now being proposed, and to illustrate the use of the method.

It will be assumed that for the conditions prevailing in a particular department, there are two sets of published data which would seem to be appropriate and between which a choice has to be made. The discordance between the two sets of data is obvious, the correct choice is not. It will be further assumed that it is required to find the exposure rate at 10 cm deep, and the steps for doing so for the two methods now in use and for the proposed procedure follows:

		Data 'A'	Data 'B'
(1)	*Basic information*		
	Backscatter factor	1.293	1.286
	Percentage depth dose 5 cm	63.3	68.7
	Percentage depth dose 10 cm	32.4	35.5
(2)	*In-air calibration*		
	Exposure rate=54 R/m		
	Surface exposure rate with backscatter	54×1.293=70	54×1.286=69.5
	Exposure rate at 10 cm deep	70×0.324=**22.7**	69.5×0.355=**24.6**
(3)	*Surface dose with full backscatter calibration*		
	Exposure rate=70 R/m		
	Exposure rate at 10 cm deep	70×0.324=**22.7**	70×0.355=**24.9**
(4)	*Using proposed procedure*		
	Exposure rate at 5 cm deep=44.5 R/m		
	Surface exposure rate with backscatter	44.5/0.633=70.4	44.5/0.687=64.8
	Exposure rate at 10 cm	70.4×0.324=**22.8**	64.8×0.355=**23.0**
	In air exposure rate	70.4/1.293=54.4	64.8/1.286=50.4

Thus, when the equipment is calibrated in air, there is a difference of 8 percent in the estimated exposure rate at 10 cm deep depending on whether Data A or Data B are used. With the surface exposure rate with full backscatter method the difference is 9 percent, whereas with the proposed method the difference is only 1 percent.

Thus, depth dose data showing considerable discordance give results in *excellent agreement at the therapeutically important depths if the proposed technique is used.* The price of this is uncertainty at the surface. However, in treatments of the type being considered, surface doses are relatively unimportant, and uncertainties of as much as 10 percent are acceptable there, though they should not be accepted in the target volume.

D. Depth Dose Calibration Methods

The phantom recommended as suitable is one either of water or of a tissue-equivalent material such as "Mix D" 5] and should be 30 x 30 cm in cross-sectional area and at least 15 cm, and preferably 20 cm deep. "Mix D" is available in sheets of the required area and about 1 cm thick. It is, however, expensive and the machining of surfaces and the cutting of a groove to take the measuring instrument may be beyond the resources of some departments. Water, on the other hand, has the overwhelming advantage of being universally standard. The major disadvantage to its use is that unless the dosimeter is placed in a watertight container serious damage to it may result. This disadvantage can, however, be overcome relatively easily by the use of a Perspex (Lucite or Plexiglas) tank to hold the water. A thin-walled Perspex tube is sealed across the tank at a suitable depth, as shown in figure IV.5, to provide a waterproof sheath for the dosimeter and for the very necessary space filling material which must be used to exclude air from around the chamber. A line can be scribed on the sides of the tank, as indicated in figure IV.5, at 5 cm above the level of the center of the tank to ensure accuracy of water depth. Such a tank is not difficult to make and may well be available commercially. Its use as a part of calibration apparatus is recommended.

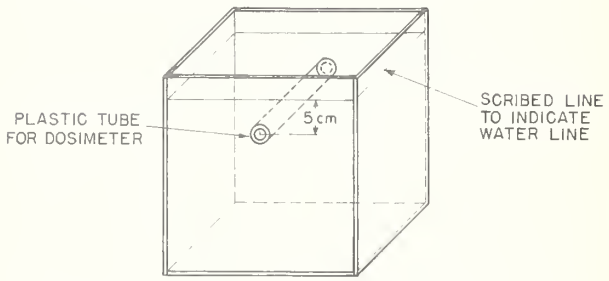

FIGURE IV.5. *Suggested plastic tank for water phantom.*

E. Radiations Above 400 kv

The foregoing was written with particular reference to radiations generated at below 400 kv, principally because it is in that energy range that most disagreement between published percentage depth dose values occurs. However, the general principle that more accurate assessments can be made if the calibration measurements are made well below the surface still holds good for higher energy radiations. In fact, one of the reasons why there is greater agreement between percentage depth dose values for the higher energy radiations may be because the reference point is not at the surface but at the peak of the "build up" curve. There are, however, one or two points of difference between the lower energy radiations and those generated at above 400 kv that warrant special mention.

The main point concerns the inapplicability of exposure for radiations of energy above a few Mev and the fact that for the higher energy radiations the central percentage depth dose values and isodose curves are for absorbed dose only, whereas for below 400 kv they represent both exposure and absorbed dose. Thus, for radiations above 400 kv, the proposed new calibration method might be described as follows:

1. For each field size measure the absorbed dose-rate at 5 cm deep in a suitable phantom, or at a depth greater than the depth of the peak of the absorbed dose "build up" curve, if that exceeds 5 cm.

2. As before, any modern depth dose tables appropriate to the radiation conditions may be used.

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3. From the measurement and the stated percentage depth dose for the point at which the measurement was made, the absorbed dose-rate at the peak of the build up curve can be calculated. This is the dose rate value usually used in treatment specifications and calculations.

It should be noted that this method of calibration gives no information about dose in the region between the surface and the point beneath the surface where the dose is a maximum—the build up region. Here, the absorbed dose will depend not only on the x-ray photon flux but also on electrons arising from extraneous objects like beam defining diaphragms, applicators, filters, etc., which may be close to the surface. The effect of this "electron contamination" of the beam will be to increase (above its level for an uncontaminated beam) the absorbed dose at all points from the skin to the peak dose depth and even a few millimeters beyond, in the manner illustrated in figure IV.6. If, for radiation up to 6 Mv, the distance of any irradiated material from the skin is of the order of 15 to 20 cm, this contamination is reduced to a minimum [6]. Should it be necessary to have solid irradiated objects closer to the surface than this, it should be remembered that materials of medium atomic number (such as copper, brass, iron, etc.) contribute less electrons to the beam than do materials of lower or higher atomic numbers. A fuller discussion of this matter may be found in the ICRU Report 10b (1962). It should be noted that the contamination also tends to increase the peak dose, though probably by not more than a percent even for quite marked contamination. To this extent the suggested calibration procedure will be in error in its estimate of the peak dose. Its estimates of deeper doses will still be accurate. A peak dose measurement, on the other hand, will include the contamination contribution but estimates of all deeper doses based on published "clean beam" depth dose data will be slightly in error.

The normal method of measuring the absorbed dose rate will be to obtain a reading with the same sort of dosimeter that has been used for the lower energy radiations, but having a calibration correction factor from the standardizing laboratory for the energy being used, or one as close as possible to it. Some laboratories only supply calibrations up to 2 Mv or to Cobalt-60 gamma-rays. However, it is reasonable to assume [7] that the factor will remain sensibly unchanged for higher energies.

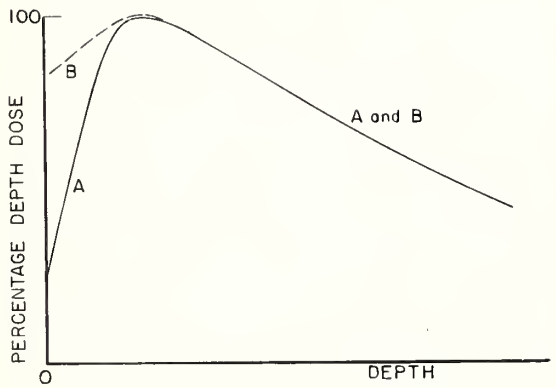

FIGURE IV.6. *Effect of electron "contamination" of high energy photon beam.*

Curve A—"clean" beam, Curve B "contaminated with electrons".

The dosimeter reading multiplied by the correction factor indicates the exposure *in the absence of the chamber* at the point upon which it is centered. That is, the factor allows for the inevitable attenuation of the beam in the "build up cap" of the instrument. When the chamber is used to determine the exposure in a phantom, as a step in the determination of the absorbed dose, some of the medium is displaced and allowance must be made for this. The precise value of this allowance has been the subject of some discussion [8, 9, 10] and would seem to require the use of an additional multiplying factor of 0.98 or 0.99 for Co^{60} gamma-rays and 4 Mv x rays. Thus to convert the dosimeter reading into absorbed dose in rads that reading must be multiplied by the calibration factor, by the "cap-correction" factor described above, and also by the appropriate roentgen to rad conversion factor.

F. Factors to Convert Exposure to Absorbed Dose

Most of the discussion so far has been centered round exposures and only passing mention has been made of conversion factors to give absorbed doses. A table of factors by which exposures have to be multiplied to give absorbed doses in rads for the type of material in question is given in table IV.1.

TABLE IV.1. *Values of* f^a

Mev	Water	Compact bone	Muscle
	Air		
0.010	0.912	3.55	0.925
.015	.889	3.96	.917
.020	.879	4.23	.917
.030	.869	4.39	.911
.040	.879	4.14	.920
.050	.892	3.58	.926
.060	.905	2.91	.929
.080	.932	1.91	.940
.10	.949	1.46	.949
.15	.962	1.05	.956
.20	.973	0.979	.963
.30	.968	.939	.957
.40	.966	.928	.955
.50	.965	.925	.957
.60	.966	.925	.957
.80	.965	.921	.957
1.0	.965	.919	.957
1.5	.964	.921	.957
2.0	.965	.921	.955
3.0	.962	.929	.955

$^a f = 0.869 \left[\frac{(\mu_{en}/\rho) \text{ medium}}{(\mu_{en}/\rho) \text{ air}} \right]$

G. Central Axis Depth Dose Tables

The calibration technique described above brings all modern depth dose data into excellent accord as far as points below about 3 cm deep are

TABLE IV.2. *Recommended central percentage depth dose data*

Half value layer, generating voltage, or nuclide	Source of data	Beam definition
0.01 to 1.0 mm Al	Number of authors. Collected in Brit. J. Radiol. Suppl. **10** [11].	Open-ended applicators.
1.0 to 8.0 mm Al	Braestrup [12]	Diaphragms.
	Johns et al., [10, 13, 14] [a]	Do.
	Wachsmann and Dimotsis [15]	Unspecified.
0.5 to 3.0 mm Cu	Johns et al., [10, 13] [a]	Diaphragms.
	Wachsmann and Dimotsis [15]	Unspecified.
1.0 to 4.0 mm Cu	Brit. J. Radiol. Suppl. **10** [11]	Closed-ended applicators.
Caesium-137	Number of authors. Collected in Brit. J. Radiol. Suppl. **10** [11].	Diaphragms.
2 Mv	ditto	Do.
Cobalt-60	Johns et al., [6, 10] [a]	Do.
4 Mv	Number of authors. Collected in Brit. J. Radiol. Suppl. **10** [11].	Do.
8 Mv	Newbery and Bewley [16] [a]	Do.
15 Mv	Brit. J. Radiol. Suppl. **10** [11]	Do.
20–25 Mv	Shapiro et al., [17]	Do.
	Laughlin et al., [18]	Do.
30–35 Mv	Wideroe [19]	Do.

[a] Tables from these authors also appear in Brit. J. Radiol. Suppl. **10** [11].

concerned, though surface doses will still be uncertain. It also makes more accurate the use of older data, but it cannot wholly compensate for percentage depth dose errors due to the use of phantom materials which would now be regarded as unsuitable, or of ionization chambers which by modern standards were too large or whose sensitivity changed too much with radiation energy changes. In order to help people wishing to use the best available data, a Study Group meeting in Geneva in 1961 under the auspices of ICRU, WHO, and IAEA has given much consideration to the enormous amount of published data. A list drawn up at that meeting of recommended central percentage depth dose data is given in table IV.2.

H. Modifications to These Tables

Even with the extensive selection of data quoted, it will not always be possible to find values for the half value layer or radiation energy, for the SSD, or for the field sizes or shapes being used. Some modifications of, or estimates from, existing data may have to be made.

1. *HVL.* If the precise quality being used is not included in the tables, suitable information can usually be obtained by interpolation between existing data. However, it must be remembered that this problem is only likely to arise in the energy range below 400 kv, where minor quality changes are of little clinical significance. Therefore, minor adjustments of filtration or applied kilovoltage which would give a standard HVL should be seriously considered since they would not only enable standard percentage depth dose data to be used directly but also make suitable isodose curves readily available.

2. *SSD.* Conversion of values for one SSD to be suitable for another can be achieved by using one of the formula outlined by Burns [20]. These are reasonably simple, and give results accurate to about $\pm$ 2 percent but, again, the user of the less orthodox distances should consider whether the advantages they give are worth the extra trouble they cause.

3. *Field size and shape.* The recommended percentage depth dose tables mostly present data for square fields, and in the others, where the field area is stated, it is implied that the areas are square or circular. For intermediate square or circular fields, depth dose values can be obtained by simple interpolation from the tables.

Rectangular fields give smaller percentage depth doses than square or circular fields of the same area, and for each size and shape of rectangle it is possible to find an "equivalent" square (or circle) which will have a smaller area than that of the rectangular field, but the same percentage depth doses. Tables of "equivalent circles" or "equivalent rectangles" have been compiled by Day [21] and are given in tables IV.3

TABLE IV.3. *Equivalent diameters of rectangular fields*

Long axis (cm)	Short axis (cm)																							
	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	22	24	26	28	30
1	1.1																								
2	1.5	2.2																							
3	1.8	2.7	3.4																						
4	1.9	3.0	3.9	4.5																					
5	2.1	3.3	4.2	5.0	5.6																				
6	2.2	3.5	4.6	5.4	6.1	6.7																			
7	2.3	3.7	4.8	5.7	6.5	7.2	7.8																		
8	2.3	3.8	5.0	6.0	6.9	7.7	8.4	8.9																	
9	2.4	3.9	5.2	6.3	7.2	8.1	8.8	9.5	10.1																
10	2.4	4.0	5.4	6.5	7.5	8.4	9.2	9.9	10.6	11.2															
11	2.5	4.1	5.5	6.7	7.7	8.7	9.6	10.3	11.1	11.7	12.3														
12	2.5	4.2	5.6	6.8	7.9	8.9	9.9	10.7	11.5	12.2	12.8	13.4													
13	2.5	4.2	5.7	6.9	8.1	9.2	10.1	11.0	11.8	12.6	13.3	13.9	14.5												
14	2.5	4.3	5.8	7.1	8.3	9.3	10.4	11.3	12.2	13.0	13.7	14.4	15.0	15.6											
15	2.5	4.4	5.9	7.2	8.4	9.5	10.6	11.5	12.4	13.3	14.1	14.8	15.5	16.1	16.7										
16	2.6	4.4	5.9	7.3	8.5	9.7	10.7	11.8	12.7	13.6	14.4	15.2	15.9	16.6	17.2	17.8									
17	2.6	4.4	5.9	7.3	8.6	9.8	10.9	11.9	12.9	13.9	14.7	15.6	16.3	17.0	17.7	18.3	18.3								
18	2.6	4.4	6.0	7.4	8.7	9.9	11.0	12.1	13.1	14.1	15.0	15.9	16.7	17.4	18.1	18.8	19.4	20.0							
19	2.6	4.5	6.0	7.4	8.8	10.0	11.2	12.3	13.3	14.3	15.3	16.2	17.0	17.8	18.5	19.3	19.9	20.5	21.1						
20	2.6	4.5	6.1	7.5	8.8	10.1	11.3	12.4	13.5	14.5	15.5	16.4	17.3	18.1	18.9	19.7	20.4	21.0	21.6	22.2					
22	2.6	4.5	6.1	7.6	8.9	10.2	11.5	12.6	13.7	14.8	15.8	16.8	17.8	18.7	19.5	20.3	21.1	21.9	22.6	23.2	24.4				
24	2.6	4.5	6.2	7.7	9.0	10.3	11.6	12.8	13.9	15.0	16.1	17.2	18.2	19.1	20.0	20.9	21.7	22.6	23.3	24.1	25.4	26.6			
26	2.6	4.5	6.2	7.7	9.1	10.4	11.7	12.9	14.1	15.2	16.4	17.4	18.5	19.4	20.4	21.4	22.3	23.1	24.0	24.8	26.2	27.6	28.7		
28	2.7	4.6	6.2	7.7	9.1	10.5	11.8	13.0	14.2	15.4	16.5	17.6	18.7	19.7	20.8	21.7	22.7	23.6	24.5	25.4	27.0	28.4	29.8	30.9	
30	2.7	4.6	6.2	7.7	9.2	10.5	11.8	13.1	14.3	15.5	16.7	17.8	18.9	20.0	21.0	22.0	23.0	24.0	24.9	25.8	27.6	29.2	30.6	31.9	33.1

All dimensions are in cm

TABLE IV.4. *Equivalent squares of rectangular fields*

Long axis (cm)	Short axis (cm) 1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	22	24	26	28	30
1	1.0																								
2	1.4	2.0																							
3	1.6	2.4	3.0																						
4	1.7	2.7	3.4	4.0																					
5	1.8	3.0	3.8	4.5	5.0																				
6	1.9	3.1	4.1	4.8	5.5	6.0																			
7	2.0	3.3	4.3	5.1	5.8	6.5	7.0																		
8	2.1	3.4	4.5	5.4	6.2	6.9	7.5	8.0																	
9	2.1	3.5	4.6	5.6	6.5	7.2	7.9	8.5	9.0																
10	2.2	3.6	4.8	5.8	6.7	7.5	8.2	8.9	9.5	10.0															
11	2.2	3.7	4.9	5.9	6.9	7.8	8.6	9.3	9.9	10.5	11.0														
12	2.2	3.7	5.0	6.1	7.1	8.0	8.8	9.6	10.3	10.9	11.5	12.0													
13	2.2	3.8	5.1	6.2	7.2	8.2	9.1	9.9	10.6	11.3	11.9	12.5	13.0												
14	2.3	3.8	5.1	6.3	7.4	8.4	9.3	10.1	10.9	11.6	12.3	12.9	13.5	14.0											
15	2.3	3.9	5.2	6.4	7.5	8.5	9.5	10.3	11.2	11.9	12.6	13.3	13.9	14.5	15.0										
16	2.3	3.9	5.2	6.5	7.6	8.6	9.6	10.5	11.4	12.2	13.0	13.7	14.3	14.9	15.5	16.0									
17	2.3	3.9	5.3	6.5	7.7	8.8	9.8	10.7	11.6	12.4	13.2	14.0	14.7	15.3	15.9	16.5	17.0								
18	2.3	4.0	5.3	6.6	7.8	8.9	9.9	10.8	11.8	12.7	13.5	14.3	15.0	15.7	16.3	16.9	17.5	18.0							
19	2.3	4.0	5.4	6.6	7.8	8.9	10.0	11.0	11.9	12.8	13.7	14.5	15.3	16.0	16.7	17.3	17.9	18.5	19.0						
20	2.3	4.0	5.4	6.7	7.9	9.0	10.1	11.1	12.1	13.0	13.9	14.7	15.5	16.3	17.0	17.7	18.3	18.9	19.5	20.0					
22	2.3	4.0	5.5	6.8	8.0	9.1	10.3	11.3	12.3	13.3	14.2	15.1	16.0	16.8	17.6	18.3	19.0	19.7	20.3	20.9	22.0				
24	2.4	4.1	5.5	6.8	8.1	9.2	10.4	11.5	12.5	13.5	14.5	15.4	16.3	17.2	18.0	18.8	19.6	20.3	21.0	21.7	22.9	24.0			
26	2.4	4.1	5.5	6.9	8.1	9.3	10.5	11.6	12.6	13.7	14.7	15.7	16.6	17.5	18.4	19.2	20.1	20.9	21.6	22.4	23.7	24.9	26.0		
28	2.4	4.1	5.6	6.9	8.2	9.4	10.5	11.7	12.8	13.8	14.8	15.9	16.8	17.8	18.7	19.6	20.5	21.3	22.1	22.9	24.4	25.7	27.0	28.0	
30	2.4	4.1	5.6	6.9	8.2	9.4	10.6	11.7	12.8	13.9	15.0	16.0	17.0	18.0	18.9	19.9	20.8	21.7	22.5	23.3	24.9	26.4	27.7	29.0	30.0

All dimensions are in cm.

and IV.4. An example may illustrate their use. Percentage depth dose data are required for a 15x5 cm field. Table IV.4 shows that the equivalent square would have a 7.5 cm side and, therefore, depth dose values obtained by interpolation for this size would apply to a 15x5 cm field.

For fields for irregular shape conversion is much less simple. Computation methods such as those described by Clarkson [22] and many others can be used in the same way as they are used for isodose curve production (see the next section), but they are time consuming. A busy department may, therefore, prefer to approximate the irregular area being used to the nearest circle, square or rectangle, and then apply the equivalent field technique where necessary.

I. Measurements

Should these methods, however, prove inadequate, resort may finally have to be made to direct measurements. Such are the difficulties, however, in the way of making accurate depth dose measurements, except at a single point with special apparatus like the calibration tank already described, that it is recommended that they should only be undertaken by those with adequate equipment and with considerable experience of dose measurements. Since the apparatus and methods to be used in percentage depth dose determinations are essentially the same as those to be used in the production of isodose curves, the discussion of them will be held over to the next section.

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V. Isodose Charts for Single Fields and Their Uses

A. Introduction

In the last section, the distribution of dose below the surface of the phantom along the central ray was discussed. This may be used to estimate the irradiation time needed to deliver a required dose to points along the central ray within the tumor, but it gives no information concerning the dose distribution elsewhere in the tumor. To obtain this, isodose curves are required as illustrated in figure V.1 which shows a Co^{60} beam directed through the jaw towards a tonsil.

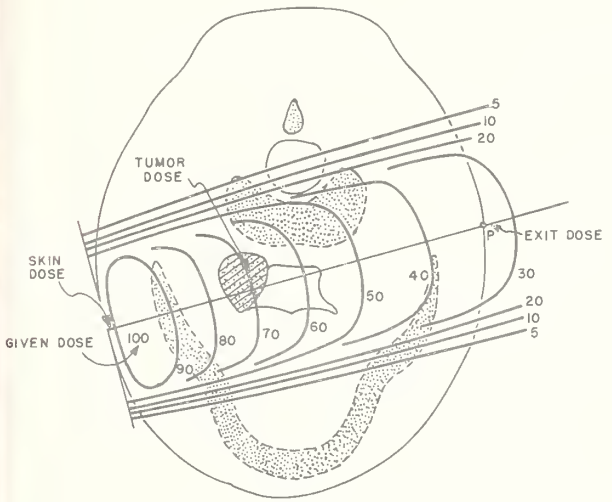

FIGURE V.1. *Isodose curves from a* Co^{60} *field superimposed on the cross section diagram of the head at the level of the tonsil.*

The isodose curves are lines of constant absorbed dose for an absorbed dose of 100 rads at the depth of the maximum. The dose variation over the particular plane section of the tumor is from 65 to 76 rads. Before attempting any radiotherapy a clinic should possess isodose charts of this kind for each of the treatment fields likely to be in regular use.

B. Acquisition of Isodose Charts

The Hospital Physicists' Association, through its "Diagrams and Data" scheme, has for many years made isodose charts from many sources generally available at nominal cost. [1] A revised catalogue of such charts and other data was prepared in 1959.

A complete set of isodose charts for closed end applicators for HVL of 1.0, 1.5, 2.0, 2.5, and 3.0 mm Cu has been published by Butterworth, London, under the sponsorship of the International Atomic Energy Agency [2]. For each quality, isodose charts for 50 different field areas are presented, full size, on translucent paper in a loose leaf book. These isodose charts are based on the axial depth dose data of the B.J.R. Supplement 10 and were calculated by Tsien [3] in collaboration with the Hospital Physicists' Association, using the Meredith and Neary method [4, 5]. The calculations were performed on the IBM computer at Temple University Medical Center, Philadelphia. In addition, this publication includes a discussion of how the isodose curves may be altered to make them applicable to open ended treatment cones. If the calibration procedure outlined in earlier sections is followed, the distinction between closed and open-ended applicators is largely unimportant. These charts cover most of the standard conditions and can be fully recommended.

For the high energy region the International Atomic Energy Agency [6] has assembled an international guide, of over 2600 entries, listing the isodose charts which are available from all parts of the world for a great variety of types and energies of x-ray and isotope machines. With this catalogue the individual user can find where to apply for isodose charts which may be most appropriate to his own particular machine. A study group of the IAEA is now preparing for publication [6] a selection of isodose curves from this larger group.

It is essential that, before any of the charts mentioned above are used in the clinic, the dose along the axis should be compared with the depth dose tables to be used in the department. In case there is disagreement, the points of intersection of the isodose curves with the axis should be adjusted to agree with the tables and corrected curves should be drawn through these points parallel to the existing ones. If the calibration procedure described in section IV is used this method will ensure accurate dose in the region of the tumour but may introduce small errors near the surface and near the edges of the beam. The modified curves will be accurate enough for most purposes and their use will certainly be better than trying to do radiotherapy with no isodose chart at all. If the depth dose data is taken from B.J.R. Supplement 10, most of the isodose curves from the sources listed above will agree exactly and no alterations will be required.

For Co^{60} and Cs^{137} units in particular, the isodose curves near the edge of the beam depend upon the penumbra and the collimating system as well as the field area and SSD. This is illustrated in figure V.2, where two isodose patterns are shown, the one on the left with a 7 mm penumbra on the skin surface and the one on the right a 31 mm penumbra at the surface. The large difference between these curves at the edge of the beam shows clearly that in selecting for use a published set of isodose curves, one should try to use those which were produced on the same type of unit with the same collimating system and source size.

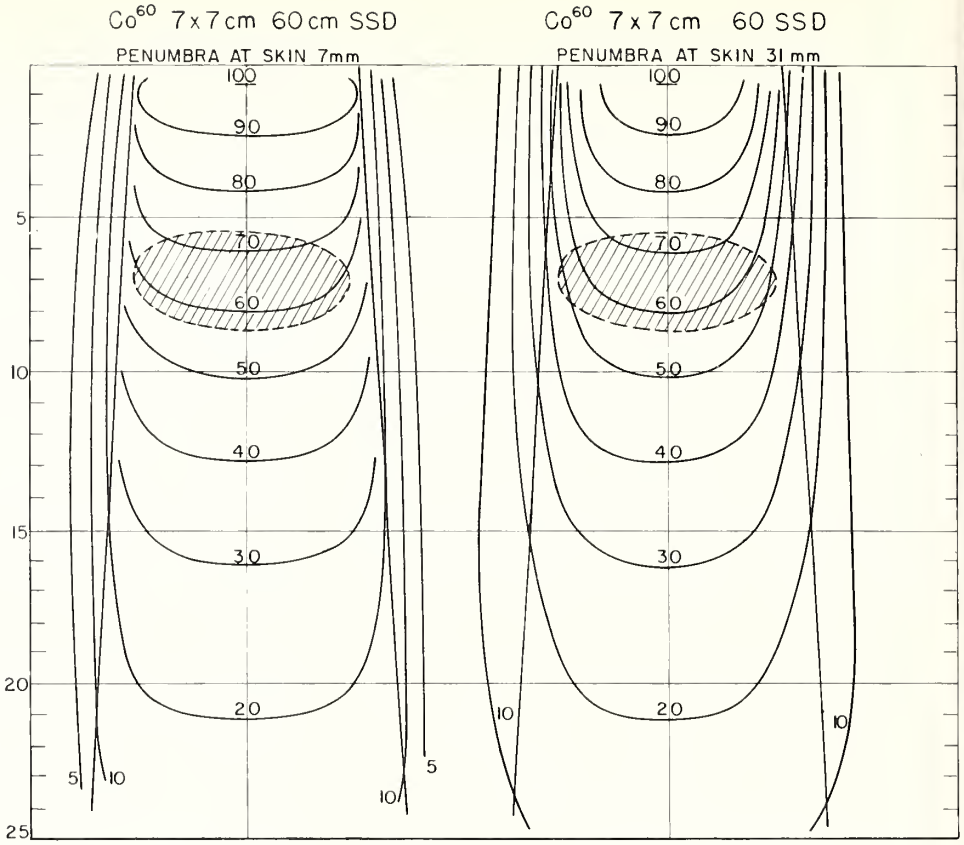

FIGURE V.2. *Isodose curves for* Co^{60}.

Left: 7x7 cm field SSD 60 cm with 7 mm penumbra.
Right: SSD 60 cm and 31 mm penumbra.

Many of the published curves giving axis depth dose data in agreement with the tables of section IV, were measured with small ionization chambers and may be relied on. If isodose charts available or in use on a particular teletherapy apparatus agree along the central axis with the data in the tables of section IV, the dose distributions near the edge of these isodose curves will probably be sufficiently accurate. If they do not agree on the central axis, adjustments to the central axis depth dose values can still be made by the procedures described above, but less confidence can be placed in the shape of the isodose curves near the edge of the beam and any checks that can be made against similar apparatus in use elsewhere would be valuable.

In figure V.2, identical sized tumors are shown on the two isodose patterns. For the beam with the small penumbra the central percentage depth dose at 7 cm deep (approximately the tumor center) is 65 percent while at the lateral boundary of the tumor it is 58 percent, a reduction of 11 percent. With the large penumbra beam the corresponding figures are 64 and 48 percent, a reduction of 24 percent. Hence, in order to achieve the same homogeneity of dose over the tumor, a wider field would have to be used in the case of the beam with the large penumbra and this would bring about the unnecessary irradiation of healthy tissues and a larger integral dose. In certain cases, involving large fields, the width of the penumbra is a small fraction of the total width and does not create a problem, but for small fields a beam with a narrow penumbra is greatly to be preferred for both fixed and moving field therapy. Fortunately with the availability today of high specific activity Co^{60}, the use of large diameter sources is disappearing with a corresponding reduction in penumbra.

These ideas are illustrated in a different way in figure V.3 which shows the dose distribution

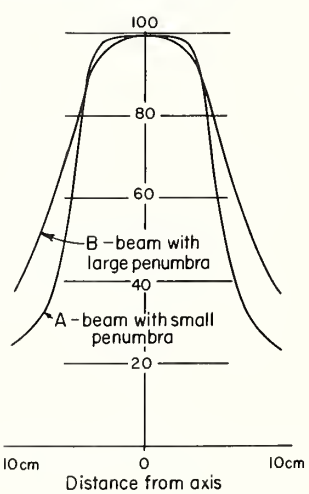

FIGURE V.3. *Relative dose distribution across a 20 cm phantom after 360° rotation using* Co^{60} *beams.*

A—penumbra 1.1 cm at the axis of rotation.
B—penumbra 4 cm at the axis of rotation.

across a cylindrical phantom of 20 cm diameter after 360 degree rotation with a 7x7 cm Co^{60} field. Curve A results from using a beam with a penumbra at the axis of rotation of 1.1 cm and curve B from a beam with a 4 cm penumbra. Distribution B gives a less uniform dose in the tumor region with much more radiation outside the tumor. The advantage of using a beam with a small penumbra is evident.

C. Construction of Isodose Curves

The production of an accurate set of isodose curves by either calculation or measurement is not an easy task.

Measurements should be made in a cubic water phantom of side at least 30 cm. Some type of device is required to move a small probe ionization chamber in a precise manner within the phantom. The probe should be of air-equivalent material, and if cylindrical in shape should have an inside diameter 5 mm or less and length 1.5 cm or less. For x-ray beams some method is required to monitor the x-ray beam so that the probe ion current may be corrected for output fluctuations. With a small chamber of this nature great care is necessary to avoid stem leakage effects while large chambers are incapable of measuring the dose at a "point"—a necessity if bad distortion in the curves is to be avoided near the edge of the beam. Since elaborate equipment of this kind is seldom available in the smaller center, it is usually quite impractical to attempt any measurements of isodose curves. Results obtained with inadequate equipment are almost certain to contain large errors and be much less accurate than isodose curves obtained by careful modification of available curves which have been properly measured.

The determination of isodose curves by calculation is not easy and it is time consuming unless some type of computer is available with trained personnel to operate it. All the methods developed to date are an extension of the one discussed originally by Clarkson [7]. In this method, the primary and secondary components of the radiation at each point in the phantom are determined separately. The scatter component is obtained from scatter functions which in turn are determined from depth dose data for circular fields. With a high speed computer, Clarkson's method can be made very precise and was the basis for many of the tables for rectangular fields in Supplement 10 [8], and those due to Aspin et al. [9] and Johns [10].

An alternative mathematical method developed by Meredith and Neary [4, 5] may also be used, or a modification of Clarkson's method, due to Tranter [11]. The last named method is probably the quickest if no computer is available, but none of these procedures is simple and they cannot be performed without trained personnel.

Since both measurement and calculation of isodose curves are skilled tasks, the user who lacks experience in this work can feel reassured that the charts recommended here are likely to be at least as accurate as any he could produce himself. The calculation method, although time consuming, will yield the same results as the standard data.

In the calculation methods mentioned above and, therefore, for any charts produced by them, it is assumed that the distribution of exposure rate across a plane normal to the central ray is circularly symmetrical about that ray. This is true for radioactive sources symmetrically placed on the central ray and is usually true for x rays produced at a transmission target. In the case of an x-ray tube with a reflection target there is usually symmetry on either side of the central ray along a line at right angles to the tube axis. Along a line parallel to the tube axis, however, the dose rate falls off more quickly in either the cathode or the anode direction depending on the target angle. Modern x-ray tubes usually have target angles chosen to achieve symmetry, but the user is advised to check that the distribution is, in fact, satisfactory and, therefore, that standard isodose charts may be used. It is also recommended that the distribution check should be repeated from time to time since "pitting" of the target may introduce asymmetry where none originally existed.

Asymmetrical distributions can be corrected by specially shaped filters. Their form, as well as the change in radiation quality across the beam, and other aspects of the problem are discussed in detail in ICRU Report 10b.

One further warning note must be sounded. Many megavolt x-ray units have built-in beam flattening filters. Isodose curves from such machines cannot be applied, with accuracy, to beams from machines that lack these filters.

D. Special Fields

There are a number of ways in which normal x- or gamma-ray beams may be altered for special purposes by using, for example, beam flattening filters and wedge filters. The effects of these filters are illustrated in figure V.4.

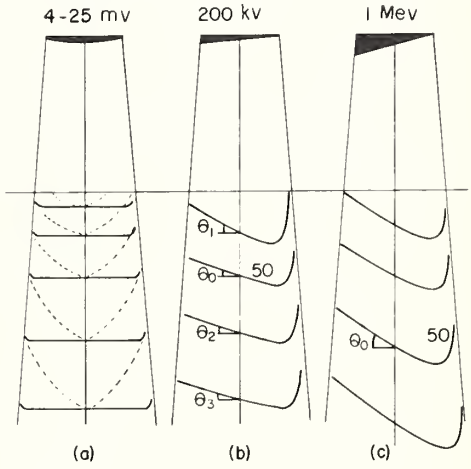

FIGURE V.4. *Effect on isodose curves on filters.*
(a) Beam flattening filter. Without such a filter the isodose curves from a high energy x-ray beam (4–25 Mv) tend to be pointed as indicated by the dotted curves. (b) and (c) Wedge filters. For 200 kv radiation the wedge angle θ tends to decrease with depth, while for the radiation of 1Mv and upwards the angle θ tends to be constant.

Beam flattening filters are used to reduce the exposure along the axis in such a way as to flatten the isodose curves out to near the edge of the beam (fig. V.4a). Filters of this kind are almost always necessary in the energy range 4 to 25 Mv. [12, 13, 14] x rays generated at these energies tend to be concentrated in the direction the high speed electrons had just before they hit the transmission target. A beam produced in this way without a compensating filter gives pointed isodose curves similar to the dotted ones of figure V.4a. In all equipment the accurate alignment in the beam of the compensating filter is important but it is absolutely essential in very high energy installations where slight errors in the positioning of the filter will markedly affect the isodose pattern. Frequent checks as to its proper alignment should be carried out.

For Co^{60} and lower energy radiation the intensity of the beam over the entrance port is essentially constant and the beam flattener is usually not required. However, in certain applications it is advantageous to flatten the curves of figure V.2 by a thin filter to give the type of curve shown in figure V.4a [12, 15, 16].

For x rays generated in the 200 kv range scattered radiation constitutes the major portion of the dose at the depth and the filter will only flatten the distribution exactly at one depth. For the use of such filters, see references 17, 18, and 19. *Wedge filters* [20] turn through an angle θ, isodose curves which normally cut the central axis at right angles. For any given filter, because of scatter, this angle decreases with increasing depth, but remains more nearly constant for high energy radiation than for the 200 kv range (figs. V.4 b and c). The wedge angle θ is arbitrarily defined as the angle θ which the 50 percent isodose makes with the normal to the axis (fig. V.4c).

Beams of this type may often facilitate the production of a homogeneous dose distribution in the target volume. This may be demonstrated by comparing the dose distributions produced by two normal fields and by two wedge filtered fields directed, with an angular separation of about 90° towards the tumor. This angle, ϕ, is often called the "hinge angle." With the normal fields a "hot spot" is produced just inside the point of intersection (A in fig. V.5a) of the adjacent edges of the beam and the dose pattern over the whole volume common to both beams is very inhomogeneous (fig. V.5a). By using two wedge fields, as shown in figure V.5b, the isodose curves from the two fields can be made approximately parallel and a homogeneous dose pattern results.

The use of wedge filters in the 200 kv range is fraught with more difficulties than in the high energy range because of the scatter differences. For instance, in using a combination of two wedge-filtered fields at right angles to produce a homogeneous field distribution giving an adequate tumour dose, the dose under the extreme thin end of the wedge may be dangerously high. This danger is virtually non-existent with high energy radiations. Without accurate positioning and angling of the beams, however, much of the value of the use of wedge filters can be lost. This accuracy can be facilitated by the use of jigs and applicators serving to ensure that the position of the patient remains constant relative to the apparatus throughout the treatment. For high energy beams it can be shown that for a two-field combination the wedge field isodose curves will be parallel if $\theta = 90 - \phi/2$, i.e., if ϕ is 90° then θ should be 45°. Theoretically this means that θ must vary continuously with ϕ but it has been shown [21] that for the production of a zone in which the dose is uniform within arbitrary limits, say ± 5 percent, it is unnecessary for the isodose curves of the two fields to be strictly parallel. For any given wedge there is a range of values of ϕ which yield clinically acceptable dose distribution. Thus, in practice, it is unnecessary to have a large number of wedge angles, one or two being sufficient to deal with most clinical situations.

Source size, SSD, and collimation all affect the resultant dose distribution and so a wedge filter designed for one machine may not be appropriate for another. Even where it is, it must certainly be mounted at the same distance from the source and used for the same SSD. Some wedge isodose curves are available on request [1] and there are many references in the literature to the design and use of wedge filters [20 to 29].

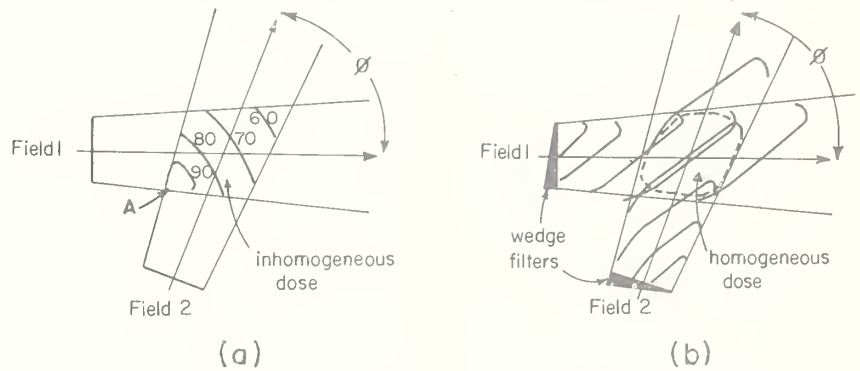

FIGURE V.5. *Comparison of dose distribution produced by two normal fields and by two wedge filter fields.*

E. Combination of Fields

For the treatment of most tumors the combination of two or more fields is required to give a chosen dose and an acceptable distribution of radiation in the target volume. The simplest combination is two fields in opposition as shown in Figure V.6 for HVL 3.0 mm Cu (left) and Co^{60} (right). In these examples the field separation is 12 cm. The individual isodose curves are shown dotted and the resultant distribution as solid lines.

For example, the summated 100 isodose line passes through the Points A, B, C, D, E, which are at the intersections of the 70 and 30, 60 and 40, 50 and 50, 40 and 60, and 30 and 70 isodose curves respectively. Figure V.6 shows that for the softer radiation, and a given dose of 100 on each field, the dose is maximum at each skin surface (120), falls along the central axis to a shallow minimum of 102 at the center. On either side of the central axis the dose is even lower, falling to 90 at the sides of the central region. On the other hand for the Co^{60} radiation the dose is essentially constant all the way along the central axis from just below each skin surface and falls off by about 10 percent at the sides.

For x rays above 2 Mv, with almost any clinically reasonable field separation, the dose is essentially constant through the block of tissue when it is irradiated by two opposing radiation fields in this way.

When more than two fields are combined, it is usually easier to obtain the resultant combination in the following way. The isodose curve drawn on a transparent sheet is laid over the contour of the body and the dose read off at a number of points which may be regularly spaced in the target volume. The isodose curve is then moved to each of the other field positions and the process repeated until the dose contributions at the selected points for all the fields have been tabulated. These are then totalled to give the resultant doses. If it is desired to give unequal doses to the different fields this may easily be taken into account by suitably weighing the tabulated values before adding the contributions. Lines of equal dose can then be plotted to give the isodose contours. The combination of three Co^{60} fields is illustrated in figure V.7. With three such fields, a tumor dose of 134 may be obtained for a given dose of 100 to each of the three fields. For high energy radiation 3 to 4 fields directed towards the tumor will usually produce a tumor dose much larger than the dose delivered to points outside the target volume. For lower energy radiations more fields are usually required to yield the same dose differential.

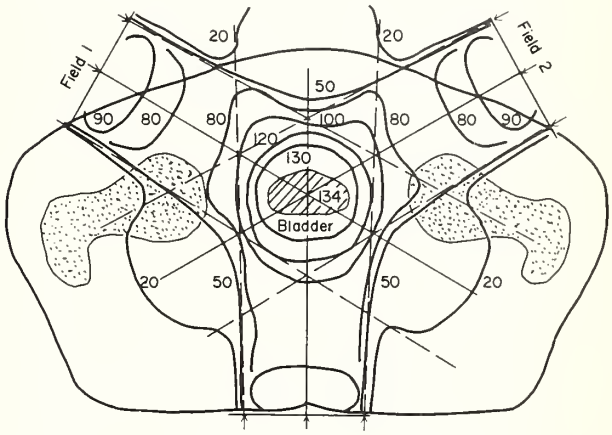

FIGURE V.7. *Isodose distribution for three 6x6 cm* Co^{60} *fields at SSD 80 cm at 120° to one another directed towards a bladder tumor.*

(NOTE: The distribution shown in figure V-7 was obtained by the summation of three isodose charts. In actual practice, isodose curves from Fields 1 and 2 will be considerably affected by the lack of bolus to correct for the oblique angle of incidence of these fields. Methods of correcting the isodose charts for oblique incidence are described in section VII.)

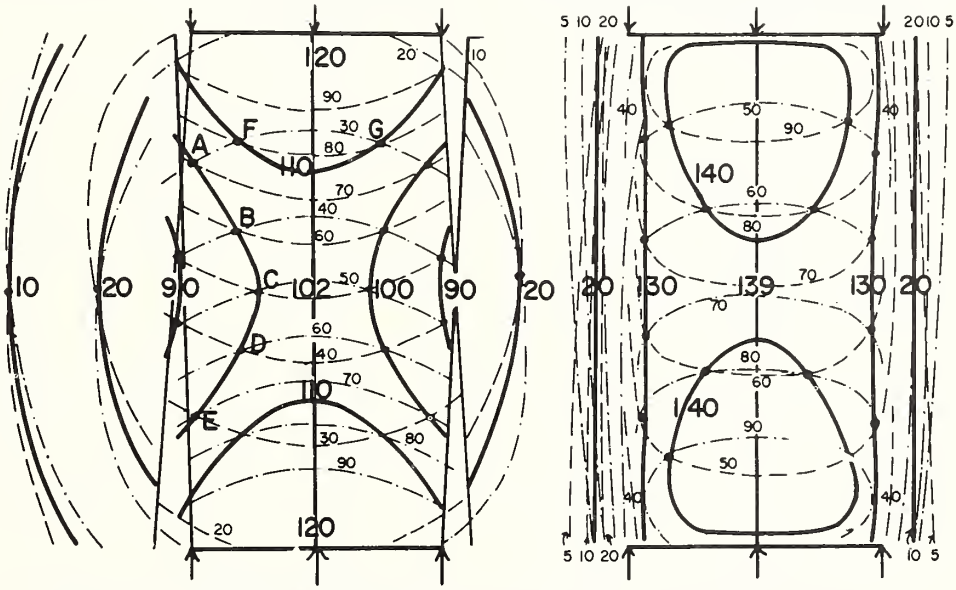

FIGURE V.6. *Distribution obtained by combining two fields in opposition, 12 cm apart.*

Left: 6x6 cm HVL 3.0 mm Cu, SSD 50 cm.
Right: 6x6 cm Co^{60}, SSD 80 cm.

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VI. Tumor Dose and Isodose Patterns for Moving Field Therapy

A. Tumor Dose Calculations

In fixed field therapy, the dose at any point in the field may be read directly from the appropriate isodose curve. Each standard isodose chart applies to a constant SSD and a constant field size at the skin surface. During moving beam therapy, both the SSD and the field size at the skin vary, so that standard isodose curves cannot be used directly in determining the dose within the patient. In rotation therapy, illustrated in figure VI.1, the only parameters which remain fixed are the *source-to-axis distance* and the *field size at the tumor.* The dose at the axis of rotation may be calculated using the concept of tissue-air ratio (TAR) which was defined originally by Johns as tumor-air ratio [1, 2, 3] (definition No. D–18, section III).

B. The Tissue-Air Ratio (TAR)

This is the ratio of the absorbed dose at the axis of rotation in the situation represented on the right hand side of figure VI.1, to the dose received there in the situation shown on the left hand side. This ratio may be expressed as:

$$\mathrm{TAR}=\frac{D_{\mathrm{air}}\cdot\left(\frac{F}{F-d}\right)^2\cdot(b)\cdot D(F-d,A,d)}{D_{\mathrm{air}}}$$

where the first factor, $\left(\frac{F}{F-d}\right)^2$ converts the dose-rate, D_{air}, at the distance F to the dose at a distance $f(=F-d)$. The second factor, b, corrects this dose for backscatter and is the backscatter factor for area $A_0(F-d)^2/F^2$. The final factor

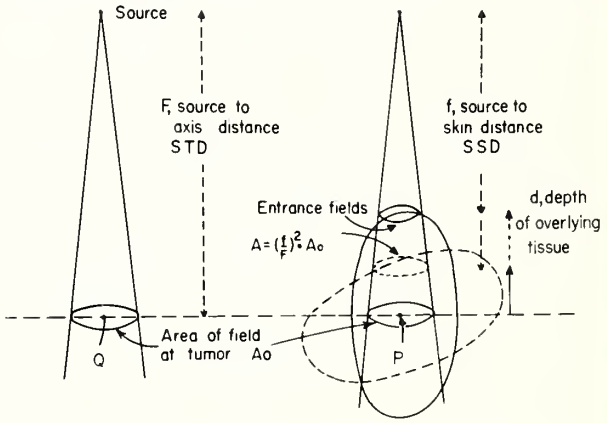

FIGURE VI.1. *Diagram to illustrate the meaning of tissue-air ratio.*

D, is the percentage depth dose appropriate to an SSD of $(F-d)$, an area $A=\frac{(F-d)^2}{F^2}A_0$ and depth d.

Although the TAR appears to be a function of the SSD it is, in fact, practically independent of F, and depends only on A_0 and d [1, 2, 3].

Tabulated values of the TAR calculated from standard depth dose tables may be found in Supplement 10 of the British Journal of Radiology (1961), and Johns [1, 2, 3, 4]. These tables give the TAR as a function of depth of overlying tissue ("d" in fig. VI.1) for a variety of rectangular and circular field sizes *measured at the axis of rotation*. They apply to all source-to-axis distances and may be used in the same way as ordinary depth dose tables. To distinguish them from percentage depth dose values, TAR values are given as decimal fractions rather than percentages. In fact, by multiplying them by 100, they can be considered as being percentage depth dose tables "referred to air" for an infinite SSD [5]. The entry in the tables for zero depth is the backscatter factor.

C. Use of Tissue-Air Ratio

The use of the tissue-air ratio will be illustrated by calculating the absorbed dose rate at the axis of rotation for the irradiation of a lung, using a 7x7 cm Co^{60} field at a source-to-axis distance of 90 cm. Suppose the exposure rate on the axis of rotation is 80 R/min and the depth "d" from the skin to the axis of rotation for the various angles of entry are as given in table VI.1. The tissue-air ratios are tabulated and averaged to give 0.507. The dose rate at the axis may now be calculated.

TABLE VI.1. *Illustrating the calculation of the dose rate at the axis of rotation during 360° rotation about a lung using a 7×7 cm Co^{60} field.*

Angle of entry	Depth (cm)	TAR [a]
0	12.0	0.592
30	13.0	.557
60	15.0	.496
90	16.5	.455
120	15.5	.482
150	13.5	.542
180	12.5	.574
210	14.0	.527
240	17.0	.442
270	18.0	.418
300	16.0	.468
330	14.0	.527
		6.080

Average TAR 0.507

[a] Taken from Brit. J. Radiol. Supplement **10**.
Exposure rate at axis=80 R/min.
Absorbed dose rate in a small sample of soft tissue placed on the axis of rotation is $80\bar{f}=76.6$ rads/min
($\bar{f}=0.957$ for Co^{60}, table IV.1)[b]

[b] f, the distance to the skin from the target, should not be confused with $\bar{f}$ which is a rad-roentgen conversion factor. Absorbed dose rate at axis in tumor $=76.6\times0.507=38.8$ rads per min.

If partial rotation is used, then only the angles of entry corresponding to this rotation should be used in the calculations. If it is desired to take into account the air cavities in the lung, the refinements discussed in section VII should be considered.

It should be noted that the TAR is defined in terms of the ratio of two absorbed doses. This means that the concept may be used for any photon energy. For radiations generated at below a few Mv, in the region where the use of the roentgen is recommended, the TAR can equally well be considered as the ratio of two exposures without introducing an error of more than a small fraction of 1 percent.

D. Rotation Isodose Distributions

The determination of complete rotation isodose distributions is a very time-consuming and difficult procedure. It may be done by considering the rotation as equivalent to irradiation by a large number of single fields and summing these fields by the methods described in section V. Before this can be done, however, one must obtain a whole set of isodose curves, each for a slightly different SSD and field-size corresponding to the SSD's actually encountered during the rotation. For example, in the rotation about a lung described in table VI.1, the depth varied from 12 to 18 cm. Hence the SSD varied from (90–12) cm to (90–18) cm, i.e., from 78 to 72 cm and the field size at the skin varied from 6.07x6.07 to 5.6x5.6 cm. Unfortunately this is about the only method which may be used for obtaining the distribution for radiation in the 200 kv range.

For higher energy radiations (above 400 kv) where scattered radiation is much less important, some simplification is possible and one set of isodose curves may be used [6]. However, even for high energy radiation, the work involved is very great and one seldom attempts to determine a complete set of isodose patterns for a routine treatment provided someone has investigated the technique for some average condition and has found that such a technique gives a good radiation distribution.

E. Dose Calculations for Points Other Than the Axis

Although a complete distribution is usually not required, one often needs the dose at a few selected points in the target volume. For this a method developed by Braestrup and Mooney [7] in which an isodose chart with curves normalized to 100 at the axis of rotation can be used. For the high energy region (about 1 Mev) the isodose pattern is continued even above the surface as indicated by figure VI.2a. Such an isodose pattern is shown superimposed over the body contour (fig. VI.2b) with its 100 percent point at the axis of rotation and the isodose pattern extended back through the skin surface. To determine the dose at a point such as Q, one lays this pattern over the body contour at an appropriate number of angles of entry (every 30° is usually satisfactory) and

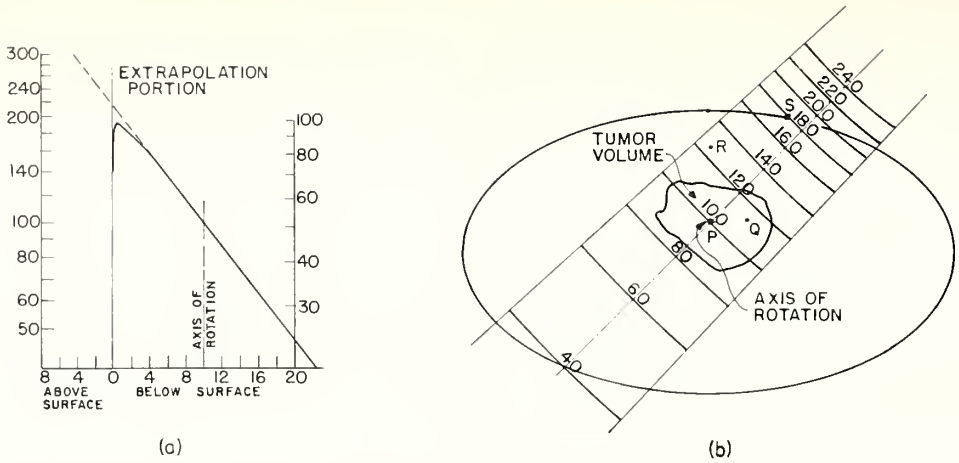

FIGURE VI.2. (a) *Depth-dose curve for 7x7 field at axis illustrating effect of normalizing to 100% at axis and extrapolating.*
(b) *How the dose at a number of points in the target volume may be estimated.*

reads off the dose from the isodose pattern for each angle of entry.

In the diagram one would read a value of 112 percent for point Q. These values are then tabulated next to the TAR values of table VI.1 for the appropriate angle of entry and multiplied by the TAR values. The corrected TAR values are then added together and averaged to give the TAR value for the point Q. This procedure may be repeated for other points of interest in the target volume. From this discussion, it is clear that one determines the dose at the point of interest (Q) by comparing it with the dose at the axis of rotation which is given the value 100. The actual dose rate at the axis is determined by the method illustrated in table VI.1.

The method described becomes increasingly inaccurate as the point of interest approaches the skin surface and should not be used for points nearer to the skin than 5 cm. However, for high energy radiation for which this method was developed [6, 7] the dose near the skin is usually only a small fraction of the dose at the tumour and need not be determined with any precision. The method would be quite satisfactory for a point such as R (deep) but not suitable for S at the skin (fig. VI.2b). The accuracy of calculation for a point such as S may be improved by using a set of isodose curves normalized to 100 at depths of 5, 10 and 15 cm as described by MacDonald [8]. At each angle of entry, one then selects the isodose curve which most nearly matches the depth of tissue for that angle of entry. This method is more precise than the former, but is slightly more difficult to use.

When rotation is carried out through 360° the maximum dose is always very near the axis and the isodose curves are essentially circular. If the beam has a small penumbra, all points which are within the beam for all angles of entry will receive about the same tumor dose in the plane traced out by the central ray, i.e., the usual plane. If the isodose curves are essentially flat, the dose to any point which is always within the beam will be very nearly the same as the dose on the axis, as illustrated in figure V.3. The larger the penumbra the less likely is this to be the case (fig. V.3) and the more desirable it is to calculate the dose at a number of points.

When only partial rotation is used, or when the rotation center is very asymmetrical, the isodose pattern is distorted as illustrated in figure VI.3 with the maximum dose shifted from the axis of rotation P to a point Q nearer the surface. By using the TAR one first determines the dose to P on the axis and then calculates, by the method just described, the dose to Q. Since the highest dose is given to Q the axis of rotation should be *displaced* away from the skin surface through which the radiation is to be delivered in order that the center of the tumour will lie at the position of maximum dose.

The complete isodose pattern in rotation therapy can be obtained experimentally. For low energy radiation (400 kv or less) small air or tissue-equivalent ion chambers may be distributed at points of interest throughout the phantom or

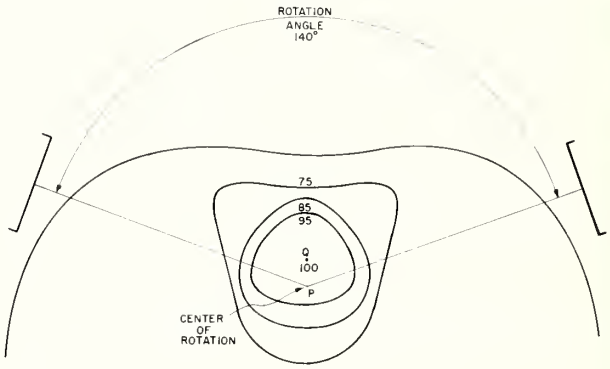

FIGURE VI.3. *Demonstration that the point of maximum dose, Q, is shifted away from the axis of rotation, P, when partial rotation is used.*

in a patient and the dose determined after one complete rotation. For high energy radiation (above 1 Mev) films or special glass rods may also be used to determine the dose. Unfortunately, neither procedure is simple and it is usually easier to estimate the dose at a few selected points by the methods described above.

F. Isodose Charts for Moving Beam Therapy

Isodose curves normalized at 100 at a depth may be obtained from the original publications [6, 7, 8, 11] and from a number of manufacturers of teletherapy equipment. Conventional charts may also be modified to this system. For instance, for equipment using a source-to-axis distance (SAD) of 60 cm, conventional charts for 50 cm SSD may be converted by considering the dose at 10 cm depth as 100 percent. The dose for point "above the surface" may be obtained by the method illustrated in figure VI.2b.

In November 1960 the International Atomic Energy Agency (IAEA) called a panel meeting to discuss among other things, isodose charts for moving beam therapy [12]. They are preparing a report in which a collection of isodose charts will be made available. To avoid any implication of clinical responsibility, all the isodose distributions will be presented in stylized form and surface contours will appear as circles, ellipses, etc., of appropriate dimensions. Co^{60} will be used as the typical energy in order to demonstrate the influence of such parameters as source-to-axis distance, length of arc, penumbral width dimensions and shape of body cross-section, etc. The effect of the use of beams of different energy will be demonstrated and all types of source movement relative to the patient in general use today will be included.

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VII. Dose Distributions in the Patient

The standard depth dose curves and isodose curves refer to a water phantom which is a cube approximately 30 cm on a side. The patient's body, however, is neither a cube nor composed wholly of water, and the isodose curves which would be obtained in a phantom built to resemble a human body will differ appreciably from the standard curves.

The aim of the therapist is to deliver to a defined volume of tissue a prescribed absorbed dose, and to be able to determine the resulting absorbed dose at any other points of interest. *One of the most important contributions to accuracy in clinical dosage is the use of high energy radiation,* for in such beams the scatter component is much less than the primary component at any point inside the beam. For instance, at a depth of 10 cm with a 10x10 cm field, the scatter component in a Cobalt-60 beam is 30 percent of the primary component, whereas in an x-ray beam with an HVL of 2.0 mm Cu, the scatter component is 250 percent of the primary component. The angular distribution of the scattered radiation is chiefly in the forward direction for high energy beams, and relatively isotropic for low energy beams. Also the penetrating power of high energy radiation is much greater than low energy radiation. Because of these differences, *the distortion of the isodose pattern produced by variations between patient and phantom in respect of shape, size and atomic composition are of much less importance with high energy than with low energy beams.*

A. Fixed Field Therapy

1. *Effect of patient size on dose distributions*

a. *Points near beam edge.* Scattered radiation reaching any point comes mainly from material placed close to the point and, therefore, when the edge of the beam lies close to the surface of the patient, there may be insufficient side-scattering material beyond the beam edge to give the full scatter and the actual dose will be less than the dose predicted by the depth dose tables. Measurements [1] have shown that for a point on the edge of the beam for field areas of 20–200 cm^2, and for radiations having half value layers ranging from 1.5 to 2.5 mm Cu, the dose is reduced by 6 percent at 5 cm depth and 12 percent for 10 cm depth. For megavolt radiation the reduction is more or less independent of depth and field size and is, at most, 5 percent.

b. *Points near exit surface.* Near the exit surface of a patient there may be insufficient

TABLE VII.1. *Factor which should be applied to the central percentage depth dose data to correct for lack of underlying tissue* [1]

(HVL 1.5 to 2.5 mm Cu; Valid depth 5–15 cm)

Depth of underlying tissue	Area of field in square centimeters						
	25	50	100	150	200	300	400
0 cm	0.81	0.76	0.72	0.70	0.68	0.67	0.66
1	.89	.86	.82	.80	.78	.76	.75
2	.93	.91	.87	.85	.84	.83	.82
3	.95	.93	.91	.89	.88	.86	.85
4	.97	.95	.93	.92	.91	.90	.89
5	.98	.97	.95	.94	.93	.93	.92
6	.99	.98	.97	.96	.96	.95	.95
7	1.00	.99	.98	.98	.98	.97	.97
8	1.00	1.00	1.00	.99	.99	.99	.99

material to give full scatter. Table VII.1 shows the corrections which should be applied for radiations having HVL's in the range from 1.5 to 2.5 mm Cu for various thickness of tissue *below* the point of interest. Thus, for example, a point anywhere from 5–15 cm below the surface and 4 cm above the exit point in a 200 cm^2 field receives 91 percent of the predicted dose. For energies above 1 Mev the corrections due to lack of backscatter are never more than 5 percent even at the exit surface.

2. *Effect of patient shape on dose estimates*

In obtaining isodose curves the beam is usually directed at right angles to the plane surface of a phantom, while during treatment of the patient the beam is often directed through a skin surface of a curved and complex form. Moreover, the central ray may be inclined at an appreciable angle to the general slope of the surface. Three consequences arise from this:

(i) Some points on the surface will be at a distance from the source greater or less than the SSD measured along the central ray;

(ii) Points below the surface, lying at the same distance from the source, may have different thicknesses of absorbing tissue between them and the source;

(iii) Points lying at the same distance from the source may be surrounded by different thicknesses of back- and side-scattering material. The errors due to this third effect are generally small, and need not be further discussed.

Errors due to the above factors may be avoided by modifying the entrance portal to correspond to the conditions for which the standard isodose curves apply. In the 200–400 kv region this is the preferred method. It may be accomplished by adding bolus to fill the air gap between skin and treatment cone as illustrated in figure VII.1 or by the use of a compression cone which may be used to force the tissue to fill the space opposite the end of the cone.

For radiations generated at above 1 Mv, the use of bolus will bring the high dose on to the skin and therefore is generally not used. For such radiations it is desirable to alter the incident radiation using a tissue compensating filter [2] as illustrated in figure VII.2, so that the dose distribution is made to approximate closely that given by the standard isodose curves at points just below the surface yet still giving build-up at the skin surface.

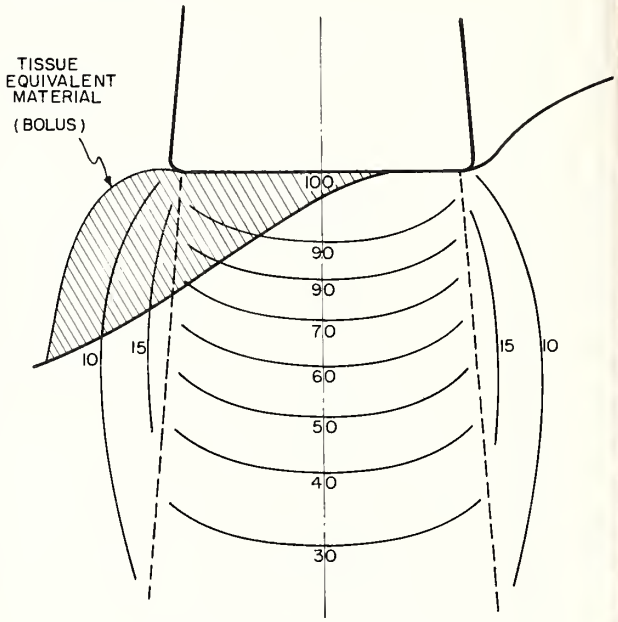

FIGURE VII.1. *The use of bolus.*

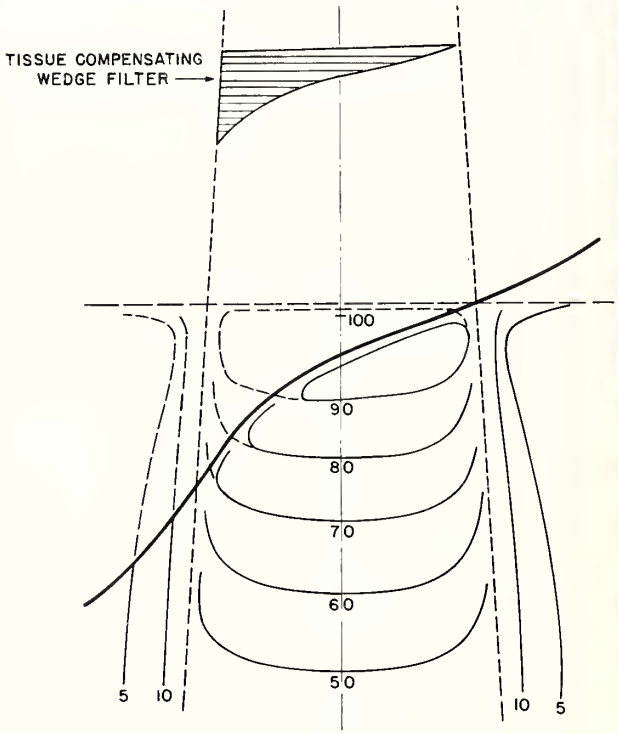

FIGURE VII.2. *How a compensating wedge filter can be used to give a distribution similar to that obtained with bolus, except in the immediate vicinity of the surface.*

For megavoltage radiations this arrangement will still provide skin sparing as indicated, provided the compensating filter is situated not less than about 15 cm from the skin.

Under some circumstances the dose distribution produced in the patient by a beam which is not normal to the surface may give a better distribution of dose than one in which bolus or a compensator is used. For these cases it is necessary to develop a system to enable one to *calculate* the correction to the dose.

The following three methods are satisfactory for angles of incidence up to about 45° for high energy beams (Co^{60} gamma rays, megavolt x rays) or up to about 30° for 200 kv beams.

a. *The Effective SSD Method.* The method is illustrated in figure VII.3 in which the surface of the patient is represented by S. It will be clear that the primary dose contribution at the point of interest, P, would be unaltered had the surface been at S′, and detailed measurements have shown that this is also practically true for the scattered radiation, especially in the megavoltage range [3]. Thus the dose at P can be regarded as being unaffected by the precise surface contour. Furthermore, since percentage depth dose values change very slowly with SSD, isodose curves appropriate to f can be used at $(f+h)$ so long as h is only a few centimeters. (The percentage depth dose at 10 cm deep changes only 0.8 percent for a change of SSD from 65 to 67 cm for Co^{60} gamma-rays). However, it must not be forgotten that the greater SSD to the point Q reduces the dose rate there, to which the percentage depth doses refer, by an inverse square law factor $f^2/(f+h)^2$.

Thus to establish the dose at P under the nonstandard conditions prevailing, the isodose curve is shifted a distance, h, as shown. The percentage depth dose at P is read off and multiplied by the inverse square law factor. In the example shown $h=1.8$ cm and $f=65$ cm. The true dose value at P is therefore $60\times\frac{65^2}{(65+1.8)^2}=57$.

b. *The Effective Attenuation Coefficient Method.* This method is illustrated in figure VII.4. The isodose pattern is placed over the body contour as indicated. The depth dose at P as read from the isodose curve is 53 percent. This dose must now be increased by a factor to allow for the lack of h cm of tissue along the line PQ. Suitable correction factors are given in table VII.2 [4]. These correction factors were obtained from the appropriate depth dose curves by removing the effect of inverse square law or from suitable tissue-air-ratio tables.

Using the appropriate corrections a value for $h=1.8$ cm gives a correction of 1.8×4.5 percent$=$ 8.1 percent. Thus the dose at P is $52+(8.1$ percent of $52)=56$ percent, in close agreement with the former value of 57 percent.

TABLE VII.2. *Factors which may be used to correct for lack of bolus (based on absorption coefficient values in Brit. J. Radiol. Suppl.* **10** [4])

Cs^{137} gamma rays	6.0% per cm.
Co^{60} gamma rays	4.5% per cm.
4 Mv X-rays	4.0% per cm.
25 Mv X-rays	2.0% per cm.

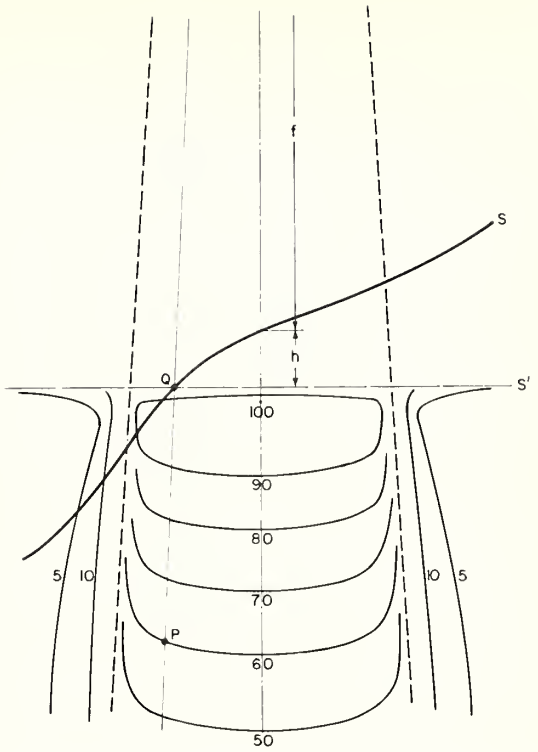

FIGURE VII.3. *The "Effective SSD Method" for estimating the dose to* P, *from a* Co^{60} *gamma-ray beam.* SSD$=$65 cm.

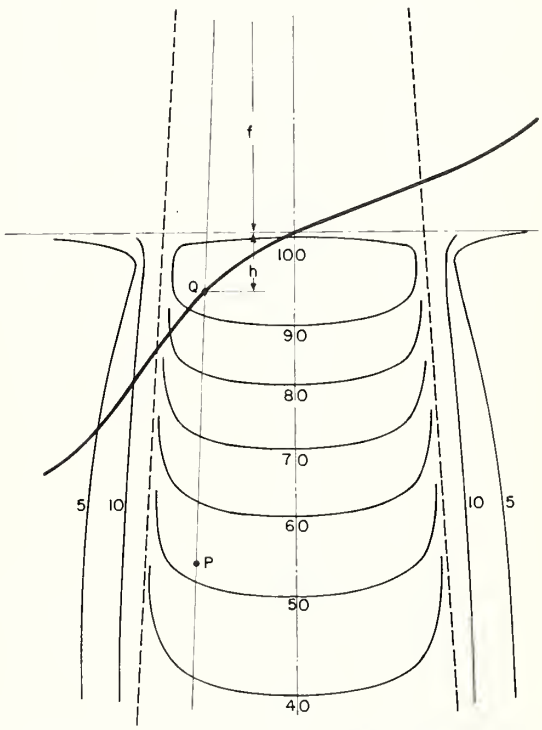

FIGURE VII.4. *The "Attenuation Method" to correct the dose at* P *for the lack of tissue of thickness* "h" *above* P. Co^{60} gamma-ray beam.

c. *The Isodose Curve Shift Method.* In this method, illustrated in figure VII.5 the isodose chart is not shifted the whole distance, h, but only some fraction of it. The percentage depth dose appropriate to the point of interest, P, is then read off directly and no further correction is required. This method was first used by Boland [5] on the 4 Mv linear accelerator and he found that the shift should be $\frac{1}{2}h$. For the Cobalt 60 machines the shift is $\frac{2}{3}h$ [6]. Figure VII.5 shows the application of this method for the same Co^{60} gamma ray beam as used in the other examples. The dose estimated at P is 56 percent which is in good agreement with the values obtained by the other methods.

Complete isodose curves can be constituted by determining the dose at a number of points using other methods.

3. *Influence of heterogeneities on absorbed dose in surrounding tissue*

The presence of tissues of different composition or density influences the dose distribution in two main ways.

(i) Since the linear attenuation coefficient in material such as bone is greater than that of soft tissue, the intensity of the beam after passing through a layer of bone is less than it would be after passing through the same thickness of soft tissues. Conversely, in lung (density, 0.25–0.4) [7, 8] the attenuation is less, and the dose beyond the lung is higher than it would be in an homogeneous body of unit density. In both these respects the effect is much less pronounced when using megavoltage than when using lower energy beams.

(ii) Since part of the dose is due to scattered radiation and since the scattering is not the same in different tissues, the dose due to scattered radiation will be influenced by any body heterogeneity. This effect is most pronounced for points situated near the heterogeneity, and becomes progressively less important at more distant points. The situation is complex and very difficult to analyse for low energy beams. For high energy beams the relative importance of scattered radiation is smaller, and as a first approximation for the energies used in clinical work, the introduction of a positional factor is unnecessary for points more than 2 cm from the heterogeneity.

The distortion of dose distribution due to these two factors depends on the quality of the radiation, the size of the field, and the size and composition of the heterogeneities. It is possible for an actual dose to differ by 50 percent or more from that calculated on the basis of homogeneous tissue, and, therefore, it is extremely important either to make the necessary corrections, or to measure the dose at the point of interest.

a. *Effect of Bone* (see also appendix II). At 200 kv the relative reduction of the dose due to the interposition of bone varies between 10 and

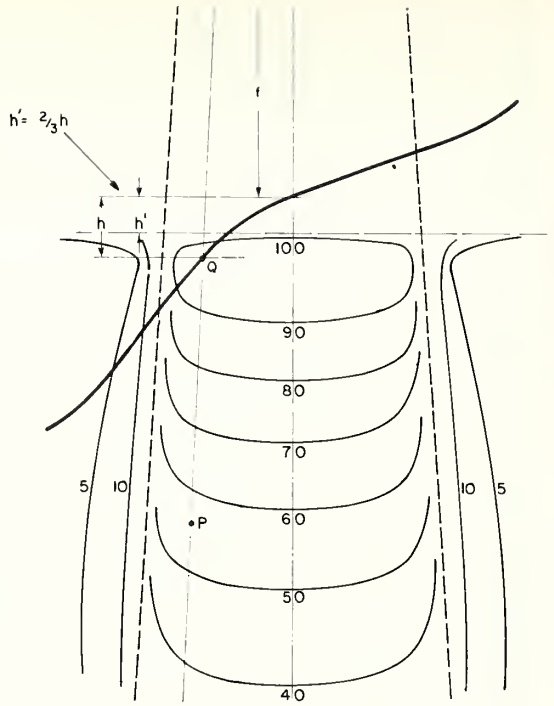

FIGURE VII.5. *The "Isodose Curve Shift Method" to determine the dose at* P. Co^{60} gamma-ray beam.

30 percent for bones of thickness varying between 1 and 3 cm. For a given quality of radiation the relative reduction of the dose varies slightly with the geometry.

For high energy (above 1 Mv) the magnitude of the shielding effect is unaffected by the chemical composition of the bone as such and, compared with an equal thickness of soft tissue is increased only in proportion to its density.

b. *Effect of Lung on Dose Distribution.* In the case of the lungs, the reduced scatter tends to reduce the dose below that given by the standard isodose curves, while the reduced attenuation in the low density lung tissue tends to increase the dose to the deeper lying tissue. Correction factors for individual fields at 200 kv as high as 2.14 or as low as 0.88 have been demonstrated in a chest-simulating phantom with no bone insertion. [9] However, the corrections necessary for the total dose when combinations of fields are used are less extreme. Corrections are negligible for a pair of anterior and posterior opposed mediastinal fields, but increase up to a factor of 1.25 for 6 symmetrically placed fields treating the mediastinum.

c. *Correction for Lung Transmission.* The simplest method of correcting for the effect of lung is to use correction factors such as those given in table VII.3, which shows the factors currently in use in Manchester [10]. For example, at 12 cm deep a particular Co^{60} beam gives a percentage depth dose of 48.1 in a water phantom. If this

had traversed 5–8 cm of lung the percentage depth dose would be increased by 20 percent to be 57.7 percent. Of necessity such corrections only apply approximately to any given patient, but if data for individual corrections are not available, the application of this sort of correction is much better than making none at all.

TABLE VII.3. *Corrections to depth dose required after the passage of a single beam through 5–8 cm of lung* [10]

300 kv	+40%
Cobalt 60	+20%
4 Mv	+15%
20 Mv	+10%

One method by which the data required for corrections on an individual may be obtained is to make measurements with a transit dosemeter (for further details see section B, Moving Field Therapy). When opposing pairs of fields are used, the corrections are accurate to within a few percent. When single fields are used it is necessary to have a knowledge, from transverse tomography, of the distributions of the various tissues in the path of the beam, if accurate use is to be made of the transit dosemeter information [12, 13]. Even so, the correction factors may be excessive for points immediately beyond lung for which lack of scatter tempers the increased transmission [10].

From transverse tomographs taken at different levels in the target volume, it is possible to estimate the amount of matter to be traversed by each pencil of the beam considered independently. This depends both on tissue contours and tissue composition. When the thicknesses of the various tissues are determined in this way, appropriate corrections may be made for tissue scatter to points in or near the lung [9].

As an alternative to making such dose corrections, it is possible in principle to compensate simultaneously at all parts of the target volume for both variations in tissue contour and variations in tissue composition, in order to obtain a uniform dose throughout the target volume. This method is illustrated in figure VII.6 which shows the distribution of tissues in one place as obtained by transverse tomography. The actual skin surface is shown as the heavy line while the equivalent skin surface (assuming unit density material and atomic number equal to that of soft tissue) is shown as the dashed line. This equivalent skin surface is valid for the radiation beam directed towards the target volume V. In positions where the equivalent skin is outside the actual skin line, the beam traverses material of density greater than 1.0 and conversely where the equivalent skin surface is inside the true skin surface, the beam traverses a region of reduced density. After the tomograph and these calculations are made, the compensating filter shown in the diagram may be constructed. The same procedure may be followed for a number of adjacent planes to yield a composite compensating filter [12].

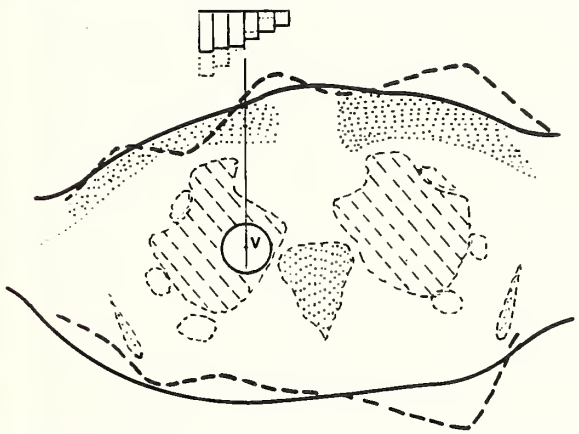

FIGURE VII.6. *The use of a compensating filter to give a uniform distribution in the target volume V.*

d. *Direct measurement with small ionization chambers.* Measurements made with a small ionization chamber or, when possible, with a probe in cavities such as the oesophagus provide an accurate measure of dose and permit a check to be made on estimates derived by other methods. When using ionization chambers inside body cavities, special care must be taken to avoid electrical leakage, and to determine the precise position of the chamber.

e. *Other air cavities.* The trachea may receive a small dose at the side located distally from the source, due to the lack of dose build-up beyond a gap [14]. Corrections to the dose are shown in table VII.4, where for example, a factor of 0.84 for a field 3 cm wide and an air gap of 4 cm. It should be emphasized that the dose will reach its maximum value at a depth of about 1 mm and due to reduced absorption in the air cavity the maximum value will be somewhat greater than given by the standard isodose curves. Since the build-up takes place in less than 1 mm the region of reduced dose is small and its clinical significance is not proven.

TABLE VII.4. *Dose reduction factors beyond an air cavity, cobalt 60 unit, 80 cm SSD, 4x4 cm field* [14]

Depth of air cavity in direction of beam	Lateral dimensions of air cavity	Dose reduction factor
1 cm	3 cm x ∞	0.97
2 cm	3 cm x ∞	.93
4 cm	3 cm x ∞	.84
4 cm	∞ x ∞	.63

B. Moving Field Therapy

In moving field therapy either the patient is rotated while the machine is held fixed or the beam is rotated and the patient held fixed. In *planar rotation* the central ray describes a plane, usually at right angles to the long axis of the patient. In *conical rotation* the central ray describes a cone, usually about a short axis of the patient. In moving field therapy the patient cannot be immobilized by the use of any treatment cone and great care must be taken to ensure that the patient

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does not shift position during the treatment. In one useful technique the operator continually keeps the tumor under observation using fluoroscopy and continually adjusts either the position of the patient's chair or the treatment field so that the tumor is continually centered in the field. This technique is particularly useful in rotation treatment of tumors in the oesophagus or pituitary.

1. *Planar Rotation.*

This type of rotation is most used in the treatment of the chest and pelvis. The irradiation of a cylindrical phantom by a field whose central axis is at right angles to the cylinder axis and which rotates about a point on that axis, produces, in the plane of rotation, isodose curves which are circles concentric with the center of rotation. In clinical practice the details of the distribution will depend on a number of factors. The following factors affect the dose distribution:

(i) *Size of patient.* Trends associated with variations in size of patient are illustrated in figure VII.7. For a given field size and the same dose at the axis of rotation, the dose delivered to the rest of the patient will be larger for a thick patient than for a thin one.

This tendency is diminished by using small fields, small penumbra, long SAD (source-to-axis-distance), and high energy beams. With a very thick patient and low energy radiation, relatively large fields may result in a uniform dose throughout the whole section of the patient, or even a dose which is larger at the periphery than at the center. 200 kv fields wider than 6 cm should not be used without great caution in treating intrathoracic lesions with complete rotation.

(ii) *Shape of patient.* Patients are rarely cylindrical. The elliptical cross section of the body tends to make the isodose curves become elliptical, with the long axis at right angles to the long axis of the body cross section (fig. VII.8). This tendency is less with high energy than with low energy beams because of the smaller attenuation of the former.

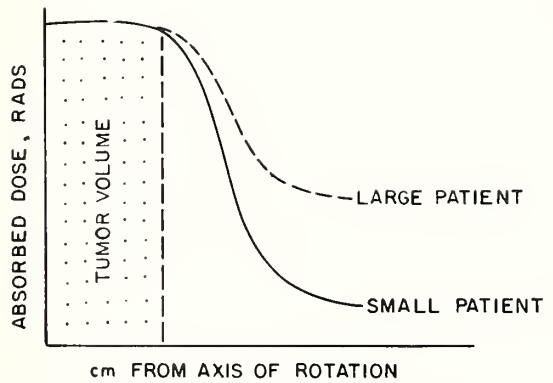

FIGURE VII.7. *How the size of the patient alters the dose distribution in full rotation.*

(iii) *Width of field.* The wider the field the greater the diameter of the high dose area. This diameter, especially for radiations up to 400 kv, is greater than the field width at the axis on account of field overlap.

(iv) *Length of field.* For 200 kv the 75 percent isodose curve makes an intercept on the axis of rotation which is roughly equal to the nominal field length, and the 90 percent isodose curve intercepts three quarters of this length for cylindrical shapes between 10 and 30 cm diameter and elliptical shapes 20×32 cm or 30×40 cm. For radiations above 1 Mev the length of the high dose region is greater than for 200 kv. For both types of radiation beam flatteners may be used to extend the high dose region still more.

(v) *Source-to-axis distance (SAD).* In general, the radiation distribution at points near the axis of rotation is insensitive to the source-to-axis distance provided the penumbra is small. This generalization is only valid for treatment volumes near the center of the patient. For Cobalt 60, where penumbra problems may be important, a rotation unit having a short source-to-axis distance may have such a large penumbra that it gives a large high-dose area outside the target volume. The shorter the SSD the less the advantage is taken of the potentialities of Cobalt 60 and the greater the dose to the non-target region. For these reasons source-to-axis distances of 75 cm or more are recommended for rotation units.

(vi) *Quality.* In general the dose distribution is insensitive to HVL over the range from 1–3 mm Cu. High energy beams give a higher ratio of tumor-dose to skin-dose and are to be preferred.

(vii) *Angle of rotation* (partial rotation). For tumors in certain positions it may be advantageous to use partial rotation. For partial rotation the high-dose region is situated nearer the proximal surface. This displacement is larger, the smaller the angle of rotation and the wider the field. To make the high-dose region coincide with the target volume, the center of the target volume must lie between the skin and axis of rotation as

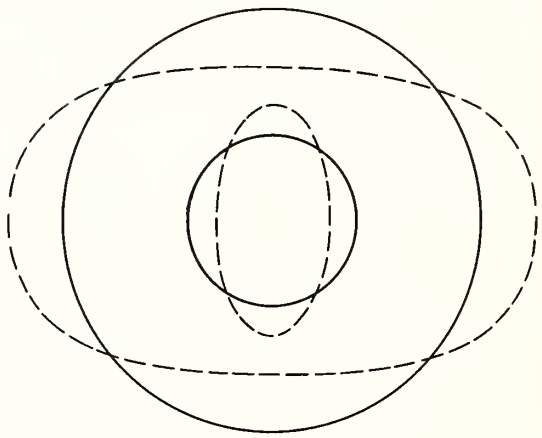

FIGURE VII.8. *The effect of body contours on isodose curve shape.*

A circular cross section leads to concentric circular curves (solid line). An elliptical section produces elliptical curves (dotted line).

illustrated in figure VII.9 and the position of the axis should be chosen accordingly.

It has been shown [15, 16] that if the isodose pattern for a particular angle of rotation and patient contour is available then this same pattern is still applicable to a larger patient of the same shape. The absorbed dose at any point will, however, be smaller by 5 percent for each cm of extra tissue. This is possible because of the small scatter from a high energy beam. By contrast, the corresponding attenuation of 200 kv x rays is about 10 to 15 percent, and is dependent on the field size so the isodose pattern is altered considerably and the method may not be used. For the same reason, at Co^{60} energies the changes in patient contour only give rise to clinically significant changes in the isodose pattern in the lesser important regions of low dosage.

2. *Conical Rotation*

When conical rotation is used, the dose distribution pattern illustrated in figure VII.10 is obtained. This shows that the high dose region is above the apex of the cone traced out by the central ray. The position of the center of the region of high dose is lowered as the field size is reduced, the energy raised and the angle widened. To obtain the distribution of figure VII.10, it is necessary to cover the surface of the patient with a cone of bolus so that the beam entering the bolus at right angles gives a situation for which the standard isodose curves apply. The dose distribution of figure VII.10 may be improved by the use of two conical rotations in opposition as illustrated in figure VII.11.

Another way to improve the dose uniformity to the target volume is to use wedge filters and conical rotation as shown in figure VII.12. One advantage of the latter arrangement is that the patient may be treated without being turned over, thus avoiding displacement of the organs in the target volume.

3. *Effects of Tissue Heterogeneity*

The presence of a region of relatively high density will cause the center of the high dose area to be displaced away from the axis of rotation, and away from the high density region (fig. VII.13). Correction factors to allow for this effect, which is more marked with low than with high energy radiation, may be obtained from measurements on phantoms, from transit dose measurements on the patient, from condenser-chamber measurements in, for example, the esophagus, or from calculations.

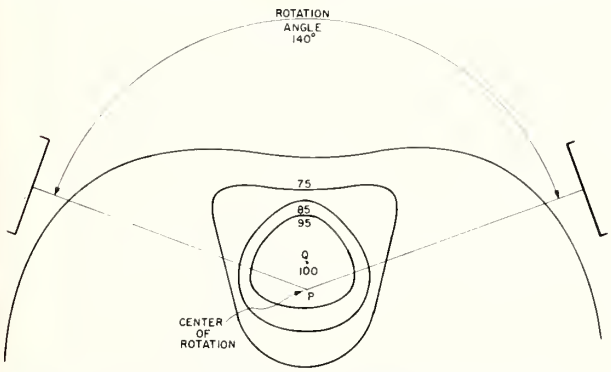

FIGURE VII.9. *The region of maximum dose at Q lies between the axis of rotation and the proximal surface.*

Transit intensity measurements [19, 20, 21] have been used in moving beam therapy to provide a simple method of correcting for average attenuation by the body during rotation. The object of transit intensity measurements is to determine the actual amount of absorbing material in the path of the beam instead of estimating this from the corresponding diameter of the body. If collimation is good, and if the distance from the body surface is sufficiently great, such measurements are free from scattered radiation, and give an accurate estimate of the average amount of material traversed, i.e., the "effective thickness" of the body. For low energy beams, correction factors of 1.2–1.8 for tumor dose have been found necessary. Collimated transit meter estimation by O'Connor [20] showed that the ratio of equivalent absorbing thickness to measured thickness was about 1.55 to 1.80, thus indicating great differences in the "equivalent outlines" of the patients, when these were traced continuously from the transit intensity measurements. Probably the most accurate transit intensity estimations are those by Fedoruk [19] which give an equivalent thickness from which the tumour dose can be computed, using tissue-air ratios for Cobalt 60.

Unless there is marked asymmetry of the absorbing masses in the chest the resultant estimate of axial dose using transit dosimeters is accurate, but if marked asymmetry does exist the dose thus calculated will apply to a point displaced from the axis, in a direction away from the absorbing masses.

Exit dose measurements [22] which do not eliminate scattered radiation give a less precise estimate of the effective patient diameter, but, with suitable precautions, are much to be preferred over the use of standard isodose charts alone.

Where possible, dose estimates, whether from transit intensity estimates or from calculations, should be checked by measurements in accessible and appropriate cavities. O'Connor [20] reported errors of 13 percent in tumor dose [22] value which had been estimated from transit intensity measurements. By using 8 Sievert chambers in a plastic catheter, placed in the esophagus, [23] the dose on the axis of the whole irradiated area may be measured and inequalities corrected by differential screening in the path of the beam (see fig. VII.14). The measured dose distribution is shown in the upper curve. When the filter is used, the uniform dose pattern shown by the lower curve is obtained.

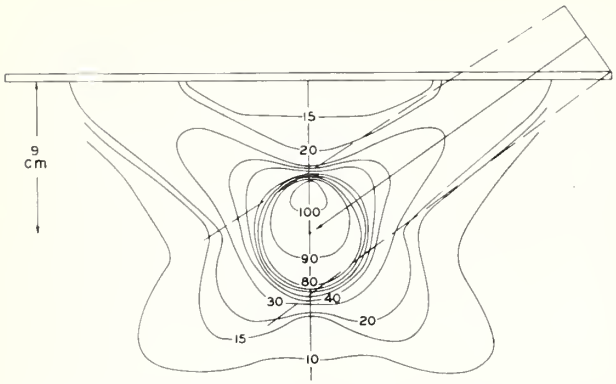

FIGURE VII.10. *Dose contours for 360°, conical rotation using a 5 cm diameter field at an SSD of 100 cm using 2 Mv; conical shaped bolus was used* [17].

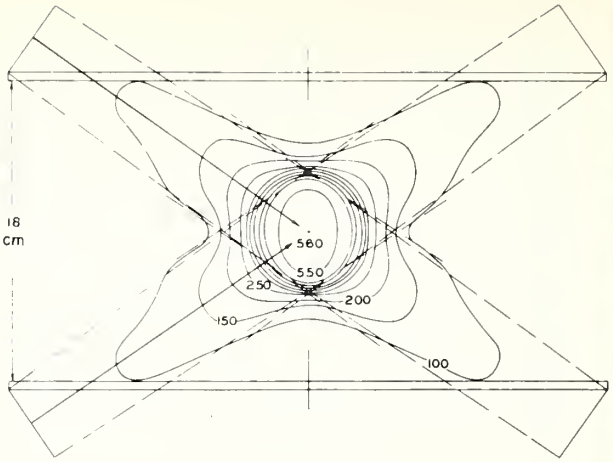

FIGURE VII.11. *Resultant distribution when two of the distributions shown in figure VII.10 are applied in opposition 18 cm apart* [17].

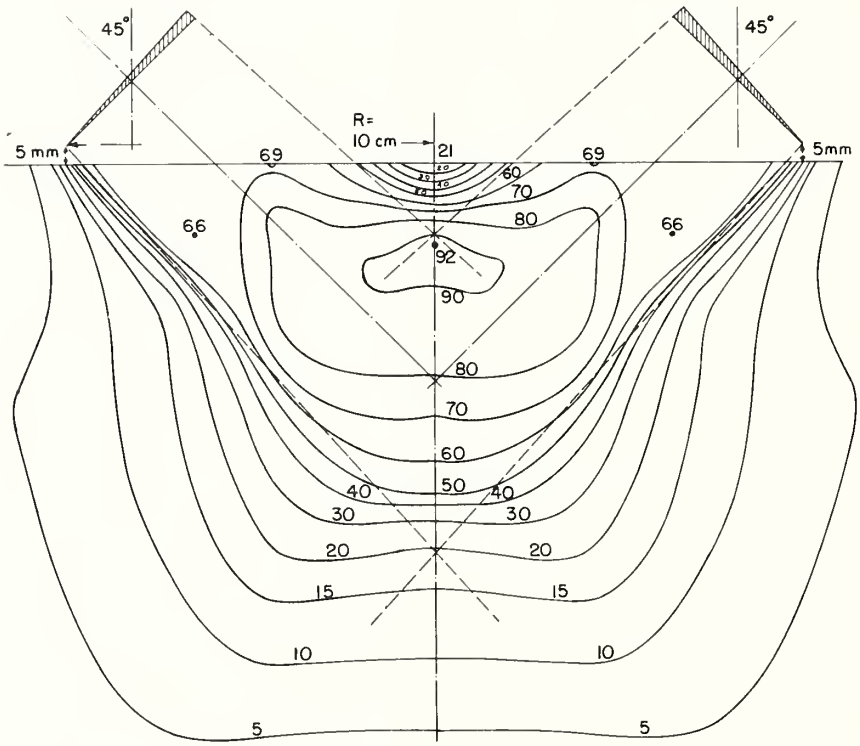

FIGURE VII.12. *Isodose distribution resulting from conical rotation with wedge filters at 250 kv* [18].

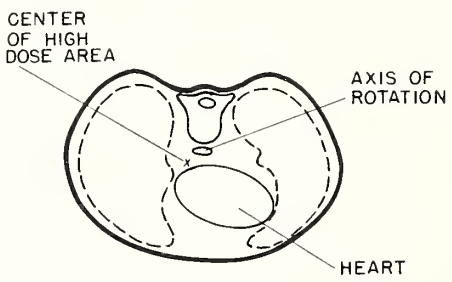

FIGURE VII.13. *How the high dose region is shifted away from the spine and axis of rotation.*

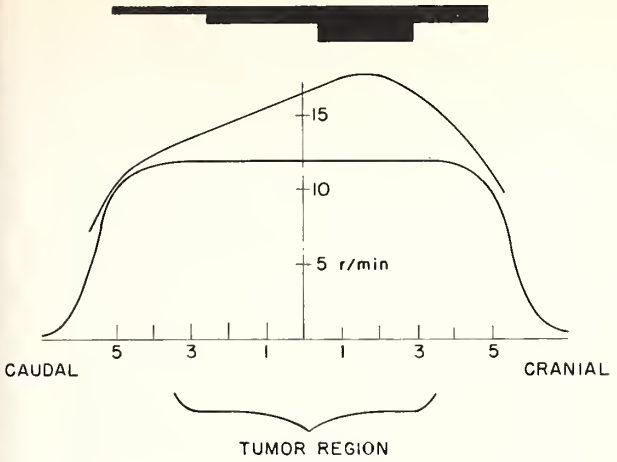

FIGURE VII.14. *The dose contour (upper curve) as obtained with Sievert chambers in the esophagus.*

When the compensating filter shown in the diagram is used, the flat response curve is obtained [23].

4. *Absorbed Dose in Compact Bone* (see appendix II)

If one neglects the inhomogeneities in the structure of bone, one can calculate the factor to convert from exposure (roentgens) to absorbed dose (rads) in bone. The change in this ratio due to scattered radiation is relatively small at the surface, and is not large even at a depth where the scatter contribution is more important. For example, an x-ray source operating at 400 kv yields a value of $\overline{f}_{\text{bone}}=1.11$ for primary radiation, while for a field of 400 cm^2 at a depth of 10 cm in a water phantom, the value of $\overline{f}$ is 1.39[24]. For x rays generated at 200 kv and having an HVL of 2.2 mm of copper, $\overline{f}_{\text{bone}}=1.32$ for the primary radiation, and has a value of 1.4 at the surface, after making allowance for back-scattered radiation. With increasing depths in the phantom the copper HVL changes from 1.2 mm at 5 cm to 1.16 at 10 cm and 0.88 at 20 cm, but the factor $\overline{f}$ only increases slowly, reaching a maximum value of about 1.9.

5. *Absorbed Dose in Soft Tissue Adjacent to Bone*

The absorbed dose in soft tissue adjacent to or in contact with the bone is of much greater clinical significance than the dose to compact bone. This tissue is irradiated by secondary electrons generated in compact bone. At the interface between bone and soft tissue there is a transition layer in which the flux of secondary electrons decreases from the level it had in compact bone to the lower level it has in soft tissue, this decrease being due to the change in atomic number, and the difference in density. The layer of soft tissue lying within range of the secondary electrons from bone receives a higher dose. The higher the energy of the radiation the smaller is the difference between the fluxes of secondary electrons in bone and in soft tissue, and the smaller is the excess irradiation near the interface. For low energy radiations, where the photoelectric effect is particularly important, the change in the electron flux in passing from bone to soft tissue is large, but since the energy of these electrons is smaller their range is shorter, and the layer of over-exposed tissue correspondingly thin.

For example, in the case of mono-energetic photons of 50 kev (corresponding to x rays emitted at a generating voltage of 130 kv), the ratio varies from 2.4 at the bone surface to 1.5 at 10 microns from the interface, and falls to unity at 25 microns (see appendix II). For x rays of higher energy, the ratio is smaller at the bone surface, but the width of the overexposed layer of tissue is greater. For example, in the case of photons of energy 100 kev (x rays emitted at a generating voltage of 250 kv) the ratio is 1.3 at the interface and decreases slowly to unity at 40 microns from the surface. Figure 3.9, appendix II indicates the distance in microns from the bone-tissue interface at which the excess dose received by the soft tissues does not exceed 10 percent. It is seen that the tissue thickness which receives a significant over-dose is a maximum for photon energies between 75 and 100 kev (x rays emitted at a generating voltage of 180 to 250 kv), but this thickness never exceeds 35 microns.

It is also important to consider the dose received by soft tissues within the Haversian systems, because this is the location of the vascular system within the bone, damage to which can lead to radiation necrosis of the bone. Figure 3.10 (appendix II) shows the variation in the dose received at the center of cavities of different dimensions for photon energies equal to 35 kev, 100 kev and 200 kev. The dose received by an osteocyte lying in a cavity 5 microns in diameter is also very high.

6. *Absorbed Dose in the Bone Marrow*

The cavities containing the bone marrow in human bones vary in diameter between 100 and 400 microns. Only the small fraction of the marrow in close proximity to the walls of these cavities will receive an overdose. The higher the x-ray energy the wider will be the transition zone where the marrow receives an overdose, but the smaller will be the difference in dose between center and walls of the cavity. One can calculate the mean absorbed dose to the marrow as a function of the photon energy and of the dimensions of the cavity. The results of such calculations are given in appendix II (fig. 3.12). One sees from this that in a cubical cavity of 400 microns to a side, the mean absorbed dose received by the marrow is some 13 percent higher than that which would be recorded if the marrow were surrounded by soft tissue, assuming a photon energy of 50 kev. At 100 kev photon energy the difference is only 7 percent, while at 200 kev photon energy it is negligible.

For further details concerning the absorbed dose in bone, references should be made in appendix II.

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VIII. Recording and Reporting in Therapy, Diagnosis and Protection

When radiation in any form is applied to biological systems, energy is transferred to tissues by the radiation, and biological effects of radiation occur. The biological effect of the radiation may be the principal objective of the application of radiation, as in radiation therapy, or it may be an undesired concomitant to the radiation, as in diagnostic studies with sources of radiation. In either case, the systematic use of the radiation requires careful measurement or calculation of the exposure to the individual and conversion of this information to data regarding the absorbed dose in tissues of interest in the biological system. Broadly speaking, these are problems within the scope of radiation physics. The next step, relating information about absorbed dose distribution in the system to its biological effect, is the problem of radiobiology.

It is convenient to group the principal factors under the following headings:

(a) The target volume.
(b) The type and energy of the radiation.
(c) The method of irradiation.
(d) The pattern of absorbed dose.
(e) The time distribution.

A. Radiation therapy

Some important reasons for keeping accurate records of radiation therapy procedures are:

(1) To ascertain that what was intended is being or has been done.

(2) To ensure that treatment information is available for the patient's management in the future.

(3) Therapy records are of medicolegal importance.

(4) To help one learn from one's own and from others' radiotherapy experience.

1. *External beam therapy.* The record form should be as simple as is consistent with completeness and flexibility. It should provide for the systematic recording of all pertinent information: identification of the patient, projected treatment plans, details of treatment, calibrated physical factors of the radiation apparatus, technique of treatment, clinical and physical daily record of treatment, and final summary of treatment. See table VIII.1.

The process of dose estimation starts with the calibration of the therapy apparatus. The final information required is the daily pattern of energy deposited by the beam of radiation at various points in the patient. These points should include at least the maximum and minimum dose in the target volume, and may also include other sites of interest, such as the skin or nearby normal structures. The process of calculating this information may be lengthy and involved. All the physical factors of the apparatus, the techniques of irradiation, the shape of the patient, and the location of the target volume within the patient must be taken into account. In complex treatment techniques, daily and cumulative exposures at several portals of entry and daily and cumulative absorbed doses at several sites of interest must be calculated and recorded. On the other hand, in some techniques (single-field superficial therapy, for example), it may be necessary to record only single portal exposure.

The outline indicated by the following table should be used as a check list in preparing reports for publication. No item should be omitted without due attention to its relevance. Many more details, in fact, may be needed in reports of complex or unusual therapeutic techniques. The outline may also be used in designing treatment record forms. It is recommended, however, that model charts be studied for guidance [1]. Such charts serve to furnish constant reminders of the pertinent data needed for future reporting as well as for immediate clinical records.

TABLE VIII.1. *External beam therapy check list*

a. *The target volume*
- (1) The shape and dimensions of the target volume.
- (2) Relation of target volume to:
  - (a) External contour of the body in the treatment position.
  - (b) Center of tumor if other than center of target volume.
  - (c) Neighboring structures of interest.
  - (d) External landmarks or reference points.

b. *The type and energy of radiation*
- (1) The type of radiation: (x rays, gamma rays, electrons, protons or neutrons).
- (2) Energy of radiation:
  - (a) For x rays: (i) up to 2 Mv; kv or Mv and HVL
    (ii) above 2 Mv; Mv only.
  - (b) For gamma-rays: the nuclide (mass number).
  - (c) For electrons and protons: the energy.
  - (d) For neutrons: the mean energy.
- (3) Output data:
  - (a) For x and gamma rays: the exposure rate at a specified point.
  - (b) For neutrons: the particle flux density or energy flux density as appropriate and at a specified point.
  - (c) Instrument and chamber or other dosimeter type.
  - (d) Monitor exposure control on apparatus.
- (4) Auxiliary information:
  - (a) X-ray tube or accelerator beam current.
  - (b) Gamma-ray source activity.

c. *The method of irradiation*
- (1) Fixed portals:
  - (a) Number of portals used and location of entry portal.
  - (b) For each portal:
    - (i) Pertinent distances to point(s) of interest, source-surface, source-site of interest (usually center of target volume), surface-site of interest;
    - (ii) dimensions and shape of field;
    - (iii) position and orientation of portal;
    - (iv) auxiliary devices: special filters or compensators;
    - (v) portal film demonstration of accuracy of beam direction.
- (2) Rotation therapy:
  - (a) Type of motion.
  - (b) Pertinent distance:
    - (i) Source to axis;
    - (ii) axis to point(s) of interest.
  - (c) Angles and arcs.
  - (d) Shape and size of portal (usually at axis).
  - (e) Auxiliary devices.

d. *The absorbed dose and its pattern*
- (1) Daily and cumulative doses in the target volume:
  - (a) Maximum.
  - (b) Minimum.
  - (c) Others when pertinent.
- (2) Doses outside target volume:
  - (a) High dose regions outside target volume.
  - (b) Doses in regions of radiosensitive tissues.
- (3) Isodose distribution in one plane, through center of target volume when feasible and pertinent. Extended to other planes, in three dimensions, when necessary.
- (4) Derivation of values:
  - (a) Source of data used for derivation of absorbed doses on basis of homogeneous medium:
    - (i) Backscatter factors;
    - (ii) depth dose or tumour-air ratio factors;
    - (iii) other corrections (field elongation, finite patient thickness, etc.);
    - (iv) exposure to absorbed dose conversion factors.
  - (b) Type, magnitude, and basis of corrections for bone, air, or other heterogeneity. Direct measurements, transit measurements, calculated measurements, calculated corrections, etc.

e. *The time distribution*
- (1) Total number of sessions (a session is a treatment or group of treatments given during one patient visit).
- (2) Total overall time and the distribution of the sessions in this time when pertinent.
- (3) Unusually high or low dose rates, unusual distributions of time factors, etc., when pertinent.

2. *Surface, intracavitary and interstitial therapy.* This type of radiation treatment is generally delivered with encapsulated radioactive sources. Often, as in interstitial radium applications, the sources of radiation are specifically arranged to fit the lesion. In other cases, such as the treatment of cancer of the uterine cervix, special applicators are used. In all cases, however, the general principles previously listed apply (see table VIII.2 for check list). In therapeutic techniques of this type, the distances involved between the sources of radiation and the sites of interest in the target volume are usually very small. Accordingly, very high dose gradients are encountered so that complete dose specification is difficult.

TABLE VIII.2. *Surface, intracavitary and interstitial therapy check list*

a. *Target volume*—see table VIII.1.
b. *Type and energy of radiation*—see table VIII.1.
c. *Method of irradiation*
  (1) Radiation sources:
    (a) Number and strength.
    (b) Active size.
    (c) Total size.
    (d) Activity per unit length.
    (e) Filtration of source capsule.
  (2) Distribution of radiation sources:
    (a) Configuration, as demonstrated by prearrangement or by roentgenograms.
    (b) Length, area, or volume of extended sources.
    (c) Applicator description.
    (d) Pattern of application, if a standard application system is used.
d. *Absorbed dose and its pattern*—see table VIII.1 Include also the specfic gamma-ray constant.
e. *Time distribution*
  (1) Half-life of the radioactive material.
  (2) Number of applications and total duration of treatment.
  (3) Number of millicurie hours.

3. *Systemic treatment with radioactive materials.* Dosimetry in this type of therapy depends on estimating the concentration ($\mu c/g$) of activity in the target volume. The amount of activity in the target volume (μc) depends on the metabolic pathways in the patient. The mass of the target volume (g) is not ordinarily known with accuracy. Accordingly, accurate dose estimation is very difficult.

The general principles of dose recording previously listed still apply, however, with some modification, as shown in table VIII.3.

TABLE VIII.3. *Systematic treatment with radioactive materials check list*

a. *Target volume* as pertinent—table VIII.1.
b. *Type and energy of radiation*—see table VIII.1.
c. (1) *Method of irradiation:*
    (a) Total activity.
    (b) Specific activity.
    (c) Amount of chemical carrier.
    (d) Chemical form and physical properties.
  (2) Mode of administration:
    (a) Oral.
    (b) Intravenous.
    (c) Other.
  (3) Deposition of the radioactivity in the biological system:
    (a) Localization:
      (i) Measurement by external means;
      (ii) estimates based on reported metabolic pathways.
d. *Absorbed dose and its pattern*, when pertinent. Include the specific gamma-ray constant.
e. *Time distribution*
  (1) Physical half-life.
  (2) Biological half-life: measurements or sources of estimation.

B. Diagnostic Uses of Radiation

In most diagnostic procedures, it is not necessary to specify absorbed dose patterns in detail, as in therapeutic procedures. However, in the development of new diagnostic methods (either with x rays or with radionuclides), detailed specification of dose should be provided so that comparison can be made between techniques and the dose at any point of interest known. Also, situations frequently arise in specific cases, such as pelvic radiography or fluoroscopy of women who do not yet know that they are pregnant, when it is necessary to have sufficient information so that doses can be computed retrospectively. In the diagnostic use of isotopes, the items listed for section A.3 (table VIII.1) apply fully. In the diagnostic use of x rays, some modifications of this list should be made, as indicated in table VIII.4.

TABLE VIII.4. *Diagnostic uses of radiation check list*

a. *Target volume*
b. *Type and energy of radiation*
c. *Method of irradiation*—see table VIII.1 when pertinent.
d. *Absorbed dose and its pattern*
e. *Time distribution*
  (1) In radiography, the milliampere-seconds for each exposure.
  (2) In fluoroscopy, the milliamperes and cumulative time of exposure.
  (3) The cumulative number of examinations and/or exposures.

Important data in this connection are given in the Adrian Committee Report [2] and in section C, which immediately follows.

C. Radiation Safety and Protection

The problems, as well as the instrumentation, in this field of radiation measurements are quite different from those previously discussed. For example there are three classes of individuals on whom dose estimates may have to be made: radiation workers, patients and the general population. Measurements too may fall into three different categories: direct measurement on the apparatus or working facilities, area monitoring and personnel monitoring. Though for more information reference should be made to publications which contain specific recommendations (3–12), the following general points should be borne in mind.

1. *Target volume*

The target volume is usually related to the maximum permissible dose recommendations i.e., whole-body, gonads, skin, extremities, or other organs of special interest.

2. *Type and energy of radiation*

The factors enumerated in table VIII.1 generally apply, though in radiation protection surveys it is also necessary to pay attention to the location of the radiation sources, details of their use (use factor, workload, occupancy factors), protective barriers, as well as to details of machine or storage safe construction.

3. *Method of irradiation*

The circumstances of irradiation vary widely. For instance in many cases it comes from radiotherapy installations, radiographic and fluoroscopic equipment or radioisotope storage or handling facilities. The general population, as well as radiation workers, may also be exposed to radiation from patients who have been treated with radioactive isotopes, or through the disposal of radioactive excreta. More limited groups may be exposed whilst engaged on radiation surveys and evaluation of hazards due to contamination in laboratories using radioactive materials, or even while handling dead patients who have recently received radioisotope treatments.

4. *Absorbed dose and its pattern*

The consideration of tables VIII.1 and VIII.3 apply.

5. *Time distribution*

The main time consideration is related to the time periods allowed for averaging in the maximum permissible dose recommendations.

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IX. Sources of Error in Clinical Dosimetry and Dose Delivery

The following paragraphs contain a list of common sources of error in clinical dosimetry and of recommended procedures for avoiding these errors. In some cases, typical values of the percentage error are quoted, but these are for the purpose of illustration only and are neither upper nor lower limits.

Errors will be discussed in the following order:

A. In connection with measurement of the radiation beam.

B. In connection with derivation of the exposure.

C. In connection with planning the treatment.

D. In connection with treating the patient.

A. Errors in Connection With Measurement of the Radiation Beam

1. *The Dosimeter.*

a. *Dosimeter Calibration.* If the dosimeter has been calibrated at a national standarizing laboratory, it will be provided with correction factors to convert its readings into roentgens. With these correction factors, measurements to within ± 2 percent of the true exposure should be possible. When the instrument has had no such calibration, greater uncertainty must be expected. Even a good quality instrument which has been handled with care may occasionally show a sudden change in sensitivity or quality dependence following slight mechanical shock. Such variations are generally more pronounced at softer qualities of radiation.

Recommendations: Because most changes in calibration are likely to arise through instrument use, the following recommendations are made to minimize errors in clinical dosimetry associated with calibration:

(i) A dosimeter called the reference chamber which is calibrated against the national standard, should be reserved solely for calibrating dosimeters used frequently in clinical dosimetry;

(ii) Such a reference chamber should be calibrated at least every two years against the national standard for all qualities at which it is to be used.

(iii) Dosimeters for regular clinical use should be calibrated against the reference chamber. Clinical dosimeters should have a check calibration against a reference chamber for the range of qualities and output in which they are used, at least twice yearly.

(iv) A constancy check against a radioactive source kept for the purpose should be performed frequently on all thimble chambers.

b. Dosimeter leakage. Dosimeter leakage may be detected as a discharge of the dosimeter with-

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out exposure to radiation. Leakage should, in the time required for the clinical measurement, produce a reading less than 1 percent of the measured dose.

Recommendation: The leakage should be corrected for or the cause eliminated.

c. Stem effects. Stem effects may be detected by the method described in section 4 below.

Recommendation: If stem effects are small a correction may be applied; if large, the dosimeter is unsuitable.

2. *Errors involved in calibration of the beam.* These may be due to:

a. Failure to use a chamber that has been calibrated for the quality being measured.

b. Failure to add the additional wall material to the chamber when required.

c. Failure to calibrate the beam under the conditions for which it will be used.

Recommendation: Measure the output for all field sizes (since there is a variation of scattered radiation with field size) and at all working distances (since scattered radiation from the collimator, tube housing, etc., will not follow the inverse square law).

d. Inaccurate positioning of the dosimeter.

Recommendation: A reproducible geometrical arrangement should be used.

e. Incorrect timing.

Recommendation: The accuracy of any timing device should be checked periodically.

f. Shutter errors and errors resulting from the time needed for voltage build-up on the x-ray tube.

Recommendation: These errors may be detected by making two exposure measurements of different duration. The difference between the two readings provides the exposure for the difference in time.

3. *Output variations.* The output of an x-ray source may change over long time intervals and may be detected by regular calibrations. The output may, however, also vary over shorter time intervals for reasons peculiar to the type of generator.

Recommendation: Output variations over hours to days can be detected by performing several measurements at different times of the daily duty cycle of the x-ray source. If output variations are significant, an exposure monitoring system should be used.

4. *Errors in the exposure monitoring system.*

a. If a parallel plate monitor is used its response, if unsealed, will depend upon the temperature.

Recommendations:

(i) Parallel plate monitoring chambers should be sealed;

(ii) Frequent calibrations of the monitor are essential;

(iii) The monitoring chamber must be calibrated for each applicator and beam quality.

b. When a thimble chamber is placed on the skin for use as a monitor, a small variation in position relative to the surface introduces a significant error.

Recommendation: The chamber should be held firmly at some position in the beam rather than being placed on the skin.

c. When used as a monitor the stem of a thimble chamber may throw a significant shadow.

Recommendation: The chamber should be so placed that no shadow is produced.

5. *Quality of the beam.*

a. Wrong measurement of HVL.

Recommendation: A technique of measuring HVL similar to that outlined in section II of this Report should be followed.

b. Inaccurate kilovoltage.

Recommendation: Correct setting of the line voltage regulator should be ensured.

B. Errors in Dose Estimation

1. *Errors in depth dose measurement.*

a. Chamber response too dependent on quality.

b. Chamber too large for accurate measurement at a point. See section IV.E.

Recommendation: For depth dose measurements, the chamber dimensions should not exceed 5 mm in diameter and 10 mm in length. For higher accuracy a displacement correction factor can be used.

c. Shadowing of scattered radiation by the stem of the ionization chamber. This effect is usually small.

d. Non-isotropic (directional) response of the chamber.

e. Inadequate size or unsuitable composition of the phantom used.

Recommendation: A phantom of tissue equivalent material, such as water, should be used. It should not be made of pressed wood or other substance whose tissue equivalence as to density and composition is uncertain.

f. Error in measuring surface exposure by thimble chamber.

Recommendation: In view of all these errors, it is recommended that depth dose measurements should be undertaken only by workers with the necessary experience and equipment.

2. *Incorrect use of depth dose data.*

a. Use of inappropriate published data.

(i) Use of published data for inappropriate S.S.D., field size or quality;

(ii) Errors in measurement of the quality of the beam;

(iii) Use of data measured under conditions producing different amounts of scattered radiation and thus introducing errors in the depth dose.

Recommendation: These errors as well as errors in surface exposure measurement should be eliminated by the adoption of the recommedned calibration procedure (section IV) which should also eliminate any serious discrepancies between the recommended sources (table IV.2).

b. Errors in connection with in-vivo condenser chamber measurements. These chambers have special problems of energy and direction dependence, susceptibility to leakage of charge due to rough handling or leakage of fluid and difficulty in knowing their position in the patient.

Recommendation: Chambers should be used in-vivo only by experienced persons.

3. *Patient characteristics.* Errors may be caused by:

a. Differences in volume between the treated region (e.g., head, neck, limb) and the phantom used to obtain the isodose curves (section VII).

b. Irregular contours of the patient causing distortion of the isodose curves (section VII).

c. Variations in atomic composition and in density (bone) occurring within the treated region (section VII, and appendix II).

C. Errors in Treatment Planning

1. *Delineation of the target volume.* Careful clinical examination and knowledge of the disease being treated are needed to avoid this error.

2. *Measurement of the patient.*

a. Inaccurate measuring instruments and poor graph-paper may result in relatively large errors in estimating and plotting patient dimensions.[4]

Recommendation: Measuring instruments, graph-paper and measurements must be carefully checked.

b. Lack of care in tranferring the patient's contour to plotting paper.

c. Variations in patient thickness along the length of the treatment field.

d. Change of cross section with position of patient (e.g., prone or supine).

e. Change of size during the course of treatment (e.g., appearance or disappearance of an effusion (ascites)).

3. *Beam direction.*

a. Movement of skin markings with change in position of patient.

b. Movement of internal organs relative to skin markings.

4. *Position of skin markings relative to target volume.*

Recommendation: The skin marking should be applied while the patient is in the treatment position. Make a radiographic check on tumor position in beam.

D. Errors in Treating the Patient

1. *Improper adjustment or operation of auxiliary devices* (e.g., light beam, back pointer, timing devices, shutter, etc.).[5]

2. *Errors in S.S.D. or in treatment position.*

3. *Incorrect type or arrangement of bolus.*

4. *Errors in sessional exposures.*

a. Output variations.

b. Calculation errors.

c. Human errors in delivery of exposure. Failure to check that all the settings of apparatus are correct. *This can lead to major errors in exposure,* (e.g., filtration, kv settings, wedge filter the wrong way round, etc.). A survey (ICRU Report 1959, section 5) in which the sessional exposures were measured by small ionization chambers on the skin, has shown that, even without mistakes in checking or calculation, about 10 percent of the patients treated received doses differing by more than 10 percent from the prescribed values.

5. *Movement of patient during treatment.*

Recommendation: Some method of immobilising the patient and of checking the patient's position during treatment should be used (e.g., observation window, closed-circuit T.V., photocell or other automatic check on field position, use of applicators and of devices to immobilise the patient).

The reduction of such remaining errors can only be achieved by careful choice of treatment technique, and by intensive training of the technical staff.

[4] Ordinary graph papers may have their engraved lines in error by as much as 10% due to humidity and temperature effects on the paper. Many inexpensive rules are also inaccurate to the same degree.

[5] In a long x-ray tube the earth's magnetic field can deflect the electron stream causing a displacement of the beam of x rays by ± 1 cm at 100 cm SSD.

Appendix I

Radiation Quantities and Units*

1. Introduction

There has recently been much discussion of the fundamental concepts and quantities employed in radiation dosimetry. This has arisen partly from the rapid increase in the number of individuals using these concepts in the expanding field of nuclear science and technology, partly because of the need for extending the concepts so that they would be of use at higher photon energies and for particulate as well as for photon radiation, but chiefly because of certain obscurities in the existing formulation of the quantities and units themselves.

The roentgen, for example, was originally defined to provide the best quantitative measure of exposure to medium energy x radiation which the measuring techniques of that day (1928) permitted. The choice of air as a standard substance was not only convenient, but also appropriate for a physical quantity which was to be correlated with the biological effect of x rays, since the effective atomic number of air is not very different from that of tissue. Thus a given biological response could be reproduced approximately by an equal exposure in roentgens for x-ray energies available at that time. Since 1928 the definition of the roentgen has been changed several times, and this has reflected some feeling of dissatisfaction with the clarity of the concept.

The most serious source of confusion was the failure to define adequately the radiation quantity of which the roentgen was said to be the unit.[1] As a consequence of this omission, the roentgen had gradually acquired a double role. The use of this name for the unit had become recognized as a way of specifying not only the magnitude but also the nature of the quantity measured. This practice conflicts with the general usage in physics, which permits, within the same field, the use of a particular unit for all quantities having the same dimensions.

Even before this, the need for accurate dosimetry of neutrons and of charged particles from accelerators or from radionuclides had compelled the International Commission on Radiological Units and Measurements (ICRU) to extend the number of concepts. It was also desired to introduce a new quantity which could be more directly correlated with the local biological and chemical effects of radiation. This quantity, *absorbed dose*, has a generality and simplicity which greatly facilitated its acceptance and in a very few years it has become widely used in every branch of radiation dosimetry.

The introduction of absorbed dose into the medical and biological field was further assisted by defining a special unit—the *rad*. One rad is approximately equal to the absorbed dose delivered when soft tissue is exposed to 1 roentgen of medium voltage x radiation. Thus in many situations of interest to medical radiology, but not in all, the numbers of roentgens and rads associated with a particular medical or biological effect are approximately equal and experience with the earlier unit could be readily transferred to the new one. Although the *rad* is merely a convenient multiple of the fundamental unit, erg/g, it has already acquired, at least in some circles, the additional connotation that the only quantity which can be measured in rads is absorbed dose. On the other hand, the rad has been used by some authors as a unit for a quantity called by them *first collision dose;* this practice is deprecated by the Commission.

Being aware of the need for preventing the emergence of different interpretations of the same quantity, or the introduction of undesirable, unrelated quantities or units in this or similar fields of measurement, the ICRU set up, during its meeting in Geneva in September 1958, an *Ad Hoc* Committee. The task of this committee was to review the fundamental concepts, quantities, and units which are required in radiation dosimetry and to recommend a system of concepts and a set of definitions which would be, as far as possible, internally consistent and of sufficient generality to cover present requirements and such future requirements as can be foreseen. The committee was instructed to pay more attention to consistency and rigor than to the historical development of the subject, and was authorized to reject any existing quantities or units which seemed to hinder a consistent and unified formulation of the concepts.

Bertrand Russell,[2] in commenting on the use and abuse of the concept of infinitesimals by mathematicians, remarks: "But mathematicians did not at first pay heed to (these) warnings. They went ahead and developed their science, and it is well that they should have done so. It is a peculiar fact about the genesis and growth of new disciplines that too much rigor too early imposed stifles the imagination and stultifies invention. A certain freedom from the strictures of sustained formality tends to promote the development of a subject in its early stages, even if this means the risk of a certain amount of error. Nonetheless, there comes a time in the development of any field when standards of rigor have to be tightened."

The purpose of the present reexamination of the concepts to be employed in radiation dosimetry

*Taken from Radiation Quantities and Units International Commission on Radiological Units and Measurements, Report 10a, National Bureau of Standards Handbook 84 (numbers refer to paragraphs in the original report).

[1] Fränz, H., and Hübner, W. Concepts and Measurement of Dose, Proceedings of Second International Conference on the Peaceful Uses of Atomic Energy, Geneva 1958, P/971 21, 101, United Nations, Geneva (1958).

[2] Russell, B., Wisdom of the West, p. 280, (Doubleday & Co., Inc., New York 1959).

was primarily "to tighten standards of rigor". If, in the process, some increased formality is required in the definitions in order to eliminate any foreseeable ambiguities, this must be accepted.

2. General Considerations

The development of the more unified presentations of quantities and units which is here proposed was stimulated and greatly assisted by mathematical models of the dosimetric field which had been proposed by some members of the committee in an effort to clarify the concepts. It appeared, however, that the essential features of the mathematical models had been incorporated into the definitions and hence the need for their exposition in this report largely disappeared. The mathematical approach is published elsewhere.[3]

As far as possible, the definitions of the various fundamental quantities given here conform to a common pattern. Complex quantities are defined in terms of the simpler quantities of which they are comprised.

The passage to a "macroscopic limit" which has to be used in defining point quantities in other fields of physics can be adapted to radiation quantities and a special discussion of this is included in the section headed "limiting procedures".

The general pattern adopted is to give a short definition and to indicate the precise meaning of any special phrase or term used by means of an explanatory note following the definition. There has been no attempt to make the list of quantities which are defined here comprehensive. Rather, the Commission has striven to clarify the fundamental dosimetric quantities and a few others (such as activity) which were specifically referred to it for discussion.

It is recognized that certain terms for which definitions are proposed here are of interest in other fields of science and that they are already variously defined elsewhere. The precise wording of the definition and even the name and symbol given to any such quantity, may at some future date require alteration if discussions with representatives of the other interested groups of scientists should lead to agreement on a common definition or symbol. Although the definitions presented here represent some degree of compromise, they are believed to meet the requirements in the field of radiation dosimetry.

3. Quantities, Units, and Their Names

The Commission is of the opinion that the definition of concepts and quantities is a fundamental matter and that the choice of units is of less importance. Ambiguity can best be avoided if the defined quantity which is being measured is specified. Nevertheless, the special units do exist in this as in many other fields. For example, the hertz is restricted, by established convention, to the measurement of vibrational frequency, and the curie, in the present recommendations, to the measurement of the activity of a quantity of a nuclide. One does not measure activity in hertz nor frequency in curies, although these quantities have the same dimensions.

It was necessary to decide whether or not to extend the use of the special dosimetric units to other more recently defined quantities having the same dimensions, to retain the existing restriction on their use to one quantity each, or to abandon the special units altogether. The Commission considers that the addition of further special units in the field of radiation dosimetry is undesirable, but continues to recognize the existing special units. It sees no objection, however, to the expression of any defined quantity in the appropriate units of a coherent physical system. Thus, to express absorbed dose in ergs per gram or joules per kilogram, exposure in coulombs per kilogram or activity in reciprocal seconds, are entirely acceptable alternatives to the use of the special units which, for historical reasons, are usually associated with these quantities.

The ICRU recommends that the use of each special unit be restricted to one quantity as follows:

The rad—solely for absorbed dose
The roentgen—solely for exposure
The curie—solely for activity.

It recommends further that those who prefer to express quantities such as absorbed dose and kerma (see below) in the same units should use units of an internationally agreed coherent system.

Several new names are proposed in the present report. When the absorbed dose concept was adopted in 1953, the Commission recognized the need for a term to distinguish it from the quantity of which the roentgen is the unit. In 1956 the Commission proposed the term *exposure* for this latter quantity. To meet objections by the ICRP, a compromise term, "exposure dose," was agreed upon.[4] While this term has come into some use since then, it has never been considered as completely satisfactory. In the meantime, the basic cause of the ICRP objection has largely disappeared since most legal codes use either the units rad or rem.

Since in this report the whole system of radiological quantities and units has come under critical review, it seemed appropriate to reconsider the 1956 decision. Numerous names were examined as a replacement for exposure dose, but there were serious objections to any which included the word dose. There appeared to be a minimum of objection to the name *exposure* and hence this term has been adopted by the Commission with the hope that the question has been permanently settled. It involves a minimum change from the older name exposure dose. Furthermore, the elimination of the term "dose" accomplishes the long-felt desire of the Commission to retain the term dose for one quantity only—the absorbed dose.

[3] Rossi, H. H., and Roesch, W. C., Field Equations in Dosimetry, Radiation Res. **16,** 783 (1962).

[4] For details, see ICRU, 1956 Report, NBS Handb. 62, p. 2 (1957).

The term "RBE dose" has in past publications of the Commission not been included in the list of definitions but was merely presented as a "recognized symbol." In its 1959 report the Commission also expressed misgivings over the utilization of the same term, *"RBE,"* in both radiobiology and radiation protection. It now recommends that the term *RBE* be used in radiobiology only and that another name be used for the linear-energy-transfer-dependent factor by which absorbed doses are to be multiplied to obtain for purposes of radiation protection a quantity that expresses on a common scale for all ionizing radiations the irradiation incurred by exposed persons. The name recommended for this factor is the *quality factor* (*QF*). Provisions for other factors are also made. Thus a *distribution factor* (*DF*) may be used to express the modification of biological effect due to non-uniform distribution of internally deposited isotopes. The product of absorbed dose and modifying factors is termed the *dose equivalent* (*DE*). As a result of discussions between ICRU and ICRP, the following formulation has been agreed upon:

The Dose Equivalent

1. For protection purposes it is useful to define a quantity which will be termed the "dose equivalent" (*DE*).
2. (*DE*) is defined as the product of absorbed dose, *D*, quality factor, (*QF*); dose distribution factor (*DF*); and other necessary modifying factors.

$$(DE)=\mathrm{D}\ \ (QF)\ \ (DF)\ \ldots\ .$$

3. The unit of dose equivalent is the "rem". The dose equivalent is numerically equal to the dose in rads multiplied by the appropriate modifying factors.

Although this statement does not cover a number of theoretical aspects (in particular the physical dimensions of some of the quantities), it fulfills the immediate requirement for an unequivocal specification of a scale that may be used for numerical expression in radiation protection.

Another new name is that for the quantity which represents the kinetic energy transferred to charged particles by the uncharged particles per unit mass of the irradiated medium. This is the same as one of the common interpretations of a concept "first collision dose," that has proved to be of great value in the dosimetry of fast neutrons. The concept is also closely related to the energy equivalent of exposure in an x-ray beam. The name proposed, *kerma*, is based on the initials of *k*inetic *e*nergy *r*eleased in *m*aterial.

Still another new name is the *energy fluence* which is here attached to the quantity in the 1953 ICRU report called *quantity of radiation*. The latter term was dropped in the 1956 ICRU report, but the concept—time integral of intensity—remains a useful one and the proposed term appears to be acceptable in other languages as well as English. A related quantity, *particle fluence*, which is equivalent to the quantity *nvt* used in neutron physics, is included to round out the system of radiation quantities.

The quantity for which the curie is the unit was referred to the committee for a name and definition. Hitherto the curie has been defined as a *quantity of the radioactive nuclide* such that $3.7\text{x}10^{10}$ disintegrations per second occur in it. However, it has never been specified what was meant by quantity of a nuclide, whether it be a number, mass, volume, etc. Meanwhile the custom has grown of identifying the number of curies of radionuclide with its transformation rate. Because of the vagueness of the original concept, because of the custom of identifying curies with transformation rate and because it appeared not to interfere with any other use of the curie, the Commission recommends that the term *activity* be used for the transformation rate, and that the curie be made its unit. It is recognized that the definition of the curie is of interest to other bodies in addition to the ICRU, but by this report we recommend that steps be taken to redefine it as $3.7\text{x}10^{10}s^{-1}$; i.e., as a unit of activity and not of quantity of a nuclide.

It is also recommended that the term *specific gamma ray constant* be used instead of *specific gamma ray emission* for the quotient of the exposure rate at a given distance by the activity. The former term focuses attention on the *constancy* of this quotient for a given nuclide rather than the *emission* of the source.

4. Detailed Considerations

A. Limiting Procedures

Except in the case of a uniform distribution of sources throughout a large region, radiation fields are in general nonuniform in space. They may also be variable in time. Many of the quantities defined in this report have to be specified as functions of space or time, and in principle they must therefore be determined for sufficiently small regions of space or intervals of time by some limiting procedure. There are conceptual difficulties in taking such limits for quantities which depend upon the discrete interactions between radiations and atoms. Similar difficulties arise with other macroscopic physical quantities such as density or temperature and they must be overcome by means of an appropriate averaging procedure.

To illustrate this procedure, we may consider the measurement of the macroscopic quantity "absorbed dose" in a nonuniform radiation field. In measuring this dose the quotient of energy by mass must be taken in an elementary volume in the medium which, on the one hand, is so small that a further reduction in its size would not appreciably change the measured value of the quotient energy by mass and, on the other hand, is still large enough to contain many interactions and be traversed by many particles.[5] If it is

[5] In interpreting radiation effects the macroscopic concept of absorbed dose may not be sufficient. Whenever the statistical fluctuations around the mean value are important, additional parameters describing the distribution of absorbed energy on a microscopic scale are necessary.

impossible to find a mass such that both these conditions are met, the dose cannot be established directly in a single measurement. It can only be deduced from multiple measurements that involve extrapolation or averaging procedures. Similar considerations apply to some of the other concepts defined below. The symbol Δ precedes the symbols for quantities that may be concerned in such averaging procedures.

In the measurement of certain material constants such as stopping power, absorption coefficient, etc., the limiting procedure can be specified more rigorously. Such constants can be determined for a given material with any desired accuracy without difficulties from statistical fluctuations. In these cases the formulae quoted in the definitions are presented as differential quotients.

B. Spectral Distributions and Mean Values

In practice many of the quantities defined below to characterize a radiation field and its interaction with matter are used for radiations having a complex energy spectrum. An important general concept in this connection is the *spectral concentration* of one quantity with respect to another. The spectral concentration is the ordinate of the distribution function of the first quantity with respect to the second. The independent quantity need not always be energy or frequency; one can speak of the spectral concentration of flux density with respect to quantum energy or of the absorbed dose with respect to linear energy transfer. The interaction constants (such as μ, S and W) referred to in this report are often mean values taken over the appropriate spectral distributions of the corresponding quantities.

C. Units

For any of the quantities defined below the appropriate unit of an internationally agreed coherent system can be used. In addition, certain special units are reserved for special quantities:

the rad for absorbed dose
the roentgen for exposure
the curie for activity.

D. Definitions

(1) *Directly ionizing particles* are charged particles (electrons, protons, α-particles, etc.) having sufficient kinetic energy to produce ionization by collision.

(2) *Indirectly ionizing particles* are uncharged particles (neutrons, photons, etc.) which can liberate directly ionizing particles or can initiate a nuclear transformation.

(3) *Ionizing radiation* is any radiation consisting of directly or indirectly ionizing particles or a mixture of both.

(4) *The energy imparted* by ionizing radiation to the matter in a volume is the difference between the sum of the energies of all the directly and indirectly ionizing particles which have entered the volume and the sum of the energies of all those which have left it, minus the energy equivalent of any increase in rest mass that took place in nuclear or elementary particle reactions within the volume.

Notes: (a) The above definition is intended to be exactly equivalent to the previous meanings given by the ICRU to "energy retained by matter and made locally available" or "energy which appears as ionization, excitation, or changes of chemical bond energies". The present formulation specifies what energy is to be included without requiring a lengthy, and possibly incomplete, catalog of the different types of energy transfer.

(b) Ultimately, most of the energy imparted will be degraded and appear as heat. Some of it, however, may appear as a change in interatomic bond energies. Moreover, during the degradation process the energy will diffuse and the distribution of heat produced may be different from the distribution of imparted energy. For these reasons the energy imparted cannot always be equated with the heat produced.

(c) The quantity *energy imparted to matter* in a given volume is identical with the quantity often called *integral absorbed dose* in that volume.

(5) The *absorbed dose* (D) is the quotient of ΔE_D by Δm, where ΔE_D is the energy imparted by ionizing radiation to the matter in a volume element, Δm is the mass of the matter in that volume element and Δ has the meaning indicated in section 4.A.

$$D=\frac{\Delta E_D}{\Delta m}$$

The special unit of absorbed dose is the *rad*.

$$1 \text{ rad}=100 \text{ erg}/g=\frac{1}{100} \text{ J}/kg$$

Note: J is the abbreviation for Joule.

(6) The *absorbed dose rate* is the quotient of ΔD by Δt, where ΔD is the increment in absorbed dose in time Δt and Δ has the meaning indicated in section 4.A.

$$\text{Absorbed dose rate}=\frac{\Delta D}{\Delta t}$$

A special unit of absorbed dose rate is any quotient of the rad by a suitable unit of time (rad/d, rad/min, rad/h, etc.).

(7) The *particle fluence*[6] *or fluence* (Φ) of particles is the quotient of ΔN by Δa, where ΔN is the number of particles which enter a sphere[7] of cross-sectional area Δa and Δ has the meaning indicated in section 4.A.

[6] This quantity is the same as the quantity, *nvt*, commonly used in neutron physics.

[7] This quantity is sometimes defined with reference to a plane of area Δa, instead of a sphere of cross-sectional area Δa. The plane is less useful for the present purposes and it will not be defined. The two quantities are equal for a unidirectional beam of particles perpendicularly incident upon the plane area.

$$\Phi=\frac{\Delta N}{\Delta a}$$

(8) The *particle flux density* or *flux density* (φ) of particles is the quotient of $\Delta\Phi$ by Δt where $\Delta\Phi$ is the particle fluence in time Δt and Δ has the meaning indicated in section 4.A.

$$\varphi=\frac{\Delta\Phi}{\Delta t}$$

NOTE: This quantity may also be referred to as particle fluence rate.

(9) The *energy fluence* (F) of particles is the quotient of ΔE_F by Δa, where ΔE_F is the sum of the energies, exclusive of rest energies, of all the particles which enter a sphere[8] of cross-sectional area Δa and Δ has the meaning indicated in section 4.A.

$$F=\frac{\Delta E_F}{\Delta a}$$

(10) The *energy flux density* or *intensity* (I) is the quotient of ΔF by Δt, where ΔF is the energy fluence in the time Δt and Δ has the meaning indicated in section 4.A.

$$I=\frac{\Delta F}{\Delta t}$$

NOTE: This quantity may also be referred to as energy fluence rate.

(11) The *kerma*[9] (K) is the quotient of ΔE_K by Δm, where ΔE_K is the sum of the initial kinetic energies of all the charged particles liberated by indirectly ionizing particles in a volume element of the specified material, Δm is the mass of the matter in that volume element and Δ has the meaning indicated in section 4.A.

$$K=\frac{\Delta E_K}{\Delta m}$$

NOTES: (a) Since ΔE_K is the sum of the initial kinetic energies of the charged particles liberated by the indirectly ionizing particles, it includes not only the kinetic energy these charged particles expend in collisions but also the energy they radiate in bremsstrahlung. The energy of any charged particles is also included when these are produced in secondary processes occurring within the volume element. Thus the energy of Auger electrons is part of ΔE_K.

(b) In actual measurements Δm should be so small that its introduction does not appreciably disturb the radiation field. This is particularly necessary if the medium for which kerma is determined is different from the ambient medium; if the disturbance is appreciable an appropriate correction must be applied.

(c) It may often be convenient to refer to a value of kerma or of kerma rate for a specified material in free space or at a point inside a different material. In such a case the value will be that which would be obtained if a small quantity of the specified material were placed at the point of interest. It is, however, permissible to make a statement such as: "The kerma for air at the point P inside a water phantom is . . .," recognizing that this is a shorthand version of the fuller description given above.

(d) A fundamental physical description of a radiation field is the intensity (energy flux density) at all relevant points. For the purpose of dosimetry, however, it may be convenient to describe the field of indirectly ionizing particles in terms of the kerma rate for a specified material. A suitable material would be air for electromagnetic radiation of moderate energies, tissue for all radiations in medicine or biology, or any relevant material for studies of radiation effects.

Kerma can also be a useful quantity in dosimetry when charged particle equilibrium exists at the position and in the material of interest, and bremsstrahlung losses are negligible. It is then equal to the absorbed dose at that point. In beams of x or gamma rays or neutrons, whose energies are moderately high, transient charged-particle equilibrium can occur; in this condition the kerma is just slightly less than the absorbed dose. At very high energies the difference becomes appreciable. In general, if the range of directly ionizing particles becomes comparable with the mean free path of the indirectly ionizing particles, no equilibrium will exist.

(12) The *kerma rate* is the quotient of ΔK by Δt, where ΔK is the increment in kerma in time Δt and Δ has the meaning indicated in section 4.A.

(13) The *exposure* (X) is the quotient of ΔQ by Δm, where ΔQ is the sum of the electrical charges on all the ions of one sign produced in air when all the electrons (negatrons and positrons), liberated by photons in a volume element of air whose mass is Δm, are completely stopped in air and Δ has the meaning indicated in section 4.A.

$$X=\frac{\Delta Q}{\Delta m}$$

The special unit of exposure is the roentgen (R).

$$1R=2.58\times10^{-4}C/kg$$ [10]

NOTES: (a) The words "charges on all the ions of one sign" should be interpreted in the mathematically absolute sense.

(b) The ionization arising from the absorption of bremsstrahlung emitted by the secondary electrons is not to be included in ΔQ. Except for this small difference, significant only at high

[8] See footnote 7.

[9] Various other methods of specifying a radiation field have been used; e.g., for a neutron source the "first collision dose" in a standard material at a specified point (see Introduction).

[10] This unit is numerically identical with the old one defined as 1 e.s.u. of charge per .001293 gram of air. C is the abbreviation for coulomb.

energies, the exposure as defined above is the ionization equivalent of the kerma in air.

(c) With present techniques it is difficult to measure exposure when the photon energies involved lie above a few Mev or below a few kev.

(d) As in the case of kerma (4D(11), note (c)), it may often be convenient to refer to a value of exposure or of exposure rate in free space or at a point inside a material different from air. In such a case the value will be that which would be determined for a small quantity of air placed at the point of interest. It is, however, permissible to make a statement such as: "The exposure at the point P inside a water phantom is . . . ."

(14) The *exposure rate* is the quotient of ΔX by Δt, where ΔX is the increment in exposure in time Δt and Δ has the meaning indicated in section 4.A.

$$\text{Exposure rate}=\frac{\Delta X}{\Delta t}$$

A special unit of exposure rate is any quotient of the roentgen by a suitable unit of time (R/s, R/min, R/h, etc.).

(15) The *mass attenuation coefficient* $\left(\frac{\mu}{\rho}\right)$ of a material for indirectly ionizing particles is the quotient of dN by the product of ρ, N, and dl, where N is the number of particles incident normally upon a layer of thickness dl and density ρ, and dN is the number of particles that experience interactions in this layer.

$$\frac{\mu}{\rho}=\frac{1}{\rho N}\frac{dN}{dl}$$

NOTES: (a) The term "interactions" refers to processes whereby the energy or direction of the indirectly ionizing particles is altered.

(b) For x or gamma radiations

$$\frac{\mu}{\rho}=\frac{\tau}{\rho}+\frac{\sigma}{\rho}+\frac{\sigma_{\text{coh}}}{\rho}+\frac{\kappa}{\rho}$$

where $\frac{\tau}{\rho}$ is the mass photoelectric attenuation coefficient, $\frac{\sigma}{\rho}$ is the total Compton mass attenuation coefficient, $\frac{\sigma_{\text{coh}}}{\rho}$ is the mass attenuation coefficient for coherent scattering, and $\frac{\kappa}{\rho}$ is the pair-production mass attenuation coefficient.

(16) The *mass energy transfer coefficient* $\left(\frac{\mu^K}{\rho}\right)$ of a material for indirectly ionizing particles is the quotient of dE_K by the product of E, ρ, and dl, where E is the sum of the energies (excluding rest energies) of the indirectly ionizing particles incident normally upon a layer of thickness dl and density ρ, and dE_K is the sum of the kinetic energies of all the charged particles liberated in this layer.

$$\frac{\mu^K}{\rho}=\frac{1}{E\rho}\frac{dE_K}{dl}$$

NOTES: (a) The relation between fluence and kerma may be written as

$$K=F\frac{\mu_K}{\rho}$$

(b) For x or gamma rays of energy $h\nu$

$$\frac{\mu_K}{\rho}=\frac{\tau_a}{\rho}+\frac{\sigma_a}{\rho}+\frac{\kappa}{\rho}$$

where

$$\frac{\tau_a}{\rho}=\frac{\tau}{\rho}\left(1-\frac{\delta}{h\nu}\right)$$

$\left(\frac{\tau}{\rho}$=the photoelectric mass attenuation coefficient, δ=average energy emitted as fluorescent radiation per photon absorbed.$\right)$ and

$$\frac{\sigma_e}{\rho}=\frac{\sigma}{\rho}\frac{E_e}{h\nu}$$

$\left(\frac{\sigma}{\rho}$=total Compton mass attenuation coefficient, E_e=average energy of the Compton electrons per scattered photon.$\right)$ and

$$\frac{\kappa_a}{\rho}=\frac{\kappa}{\rho}\left(1-\frac{2mc^2}{h\nu}\right)$$

$\left(\frac{\kappa}{\rho}$=mass attenuation coefficient for pair production, mc^2=rest energy of the electron.$\right)$

(17) The *mass energy-absorption coefficient* $\left(\frac{\mu_{en}}{\rho}\right)$ of a material for indirectly ionizing particles is $\frac{\mu_K}{\rho}(1-G)$, where G is the proportion of the energy of secondary charged particles that is lost to bremsstrahlung in the material.

NOTES: (a) When the material is air, $\frac{\mu_{en}}{\rho}$ is proportional to the quotient of exposure by fluence.

(b) $\frac{\mu_K}{\rho}$ and $\frac{\mu_{en}}{\rho}$ do not differ appreciably unless the kinetic energies of the secondary particles are comparable with or larger than their rest energy.

(18) The *mass stopping power* $\left(\frac{S}{\rho}\right)$ of a material for charged particles is the quotient of dE_s by the product of dl and ρ, where dE_s is the average energy lost by a charged particle of specified

energy in traversing a path length dl, and ρ is the density of the medium.

$$\frac{S}{\rho}=\frac{1}{\rho}\frac{dE_s}{dl}$$

NOTE: dE_s denotes energy lost due to ionization, electronic excitation and radiation. For some purposes it is desirable to consider stopping power with the exclusion of bremsstrahlung losses. In this case $\frac{S}{\rho}$ must be multiplied by an appropriate factor that is less than unity.

(19) The *linear energy transfer* (L) of charged particles in a medium is the quotient of dE_L by dl where dE_L is the average energy locally imparted to the medium by a charged particle of specified energy in traversing a distance of dl.

$$L=\frac{dE_L}{dl}$$

NOTES: (a) The term "locally imparted" may refer either to a maximum distance from the track or to a maximum value of discrete energy loss by the particle beyond which losses are no longer considered as local. In either case the limits chosen should be specified.

(b) The concept of linear energy transfer is different from that of stopping power. The former refers to energy imparted within a limited volume, the latter to loss of energy regardless of where this energy is absorbed.

(20) The *average energy* (W) *expended in a gas per ion pair formed* is the quotient of E by N_W, where N_W is the average number of ion pairs formed when a charged particle of initial energy E is completely stopped by the gas.

$$W=\frac{E}{N_W}$$

NOTES: (a) The ions arising from the absorption of bremsstrahlung emitted by the charged particles are not to be counted in N_W.

(b) In certain cases it may be necessary to consider the variation in W along the path of the particle, and a differential concept is then required, but is not specifically defined here.

(21) A *nuclide* is a species of atom having specified numbers of neutrons and protons in its nucleus.

(22) The *activity* (A) of a quantity of a radioactive nuclide is the quotient of ΔN by Δt, where ΔN is the number of nuclear transformations which occur in this quantity in time Δt and Δ has the meaning indicated in section 4.A.

$$A=\frac{\Delta N}{\Delta t}$$

The special unit of activity is the curie (c).

$$1c=3.7\times10^{10}s^{-1} \text{ (exactly)}$$

NOTE: In accordance with the former definition of the curie as a unit of quantity of a radioactive nuclide, it was customary and correct to say: "Y curies of P–32 were administered . . . ." It is still permissible to make such statements rather than use the longer form which is now correct: "A quantity of P–32 was administered whose activity was Y curies."

(23) The *specific gamma ray constant* (Γ) of a gamma-emitting nuclide is the quotient of $l^2\frac{\Delta X}{\Delta t}$ by A, where $\frac{\Delta X}{\Delta t}$ is the exposure rate at a distance l from a point source of this nuclide having an activity A and Δ has the meaning indicated in section 4.A.

$$\Gamma=\frac{l^2\Delta X}{A\Delta t}$$

Special units of specific gamma ray constant are $Rm^2h^{-1}c^{-1}$ or any convenient multiple of this.

NOTE: It is assumed that the attenuation in the source and along l is negligible. However, in the case of radium the value of Γ is determined for a filter thickness of 0.5 mm of platinum and in this case the special units are $Rm^2h^{-1}g^{-1}$ or any convenient multiple of this.

TABLE 4.1.—*Table of quantities and units*

No.	Name	Symbol	Dimensions [a]	Units		
				MKSA	egs	Special
4	Energy imparted (integral absorbed dose).	------	E	J	erg	g. rad.
5	Absorbed dose	D	EM^{-1}	J kg^{-1}	erg g^{-1}	rad.
6	Absorbed dose rate.	------	$EM^{-1}T^{-1}$	J $kg^{-1}s^{-1}$	erg $g^{-1}s^{-1}$	rad s^{-1} etc.
7	Particle fluence or fluence.	Φ	L^{-2}	m^{-2}	cm^{-2}	
8	Particle flux density.	φ	$L^{-2}T^{-1}$	$m^{-2}s^{-1}$	$cm^{-2}s^{-1}$	
9	Energy fluence	F	EL^{-2}	J m^{-2}	erg cm^{-2}	
10	Energy flux density or intensity.	I	$EL^{-2}T^{-1}$	J $m^{-2}s^{-1}$	erg $cm^{-2}s^{-1}$	
11	Kerma	K	EM^{-1}	J kg^{-1}	erg g^{-1}	
12	Kerma rate	------	$EM^{-1}T^{-1}$	J $kg^{-1}s^{-1}$	erg $g^{-1}s^{-1}$	
13	Exposure	X				R (roentgen).
14	Exposure rate	------	$QM^{-1}T^{-1}$	C $kg^{-1}s^{-1}$	esu $g^{-1}s^{-1}$	Rs^{-1} etc.
15	Mass attenuation coefficient.	$\frac{\mu}{\rho}$	L^2M^{-1}	m^2kg^{-1}	cm^2g^{-1}	
16	Mass energy transfer coefficient.	$\frac{\mu K}{\rho}$	L^2M^{-1}	m^2kg^{-1}	cm^2g^{-1}	
17	Mass energy absorption coefficient.	$\frac{\mu_{en}}{\rho}$	L^2M^{-1}	m^2kg^{-1}	cm^2g^{-1}	
18	Mass stopping power.	$\frac{S}{\rho}$	EL^2M^{-1}	J m^2kg^{-1}	erg cm^2g^{-1}.	
19	Linear energy transfer.	L	EL^{-1}	J m^{-1}	erg cm^{-1}.	kev(μm)$^{-1}$.
20	Average energy per ion pair.	W	E	J	erg	ev.
22	Activity	A	T^{-1}	s^{-1}	s^{-1}	c(curie).
23	Specific gamma-ray constant.	Γ	QL^2M^{-1}	C m^2kg^{-1}	esu cm^2g^{-1}.	$Rm^2h^{-1}c^{-1}$ etc.
	Dose equivalent.	DE				rem

[a] It was desired to present only one set of dimensions for each quantity, a set that would be suitable in both the MKSA and electrostatic-cgs systems. To do this it was necessary to use a dimension Q, for the electrical charge, that is not a fundamental dimension in either system. In the MKSA system (fundamental dimensions M, L, T, I) Q represents the product IT; in the electrostatic-cgs system (M, L, T) it represents $M^{1/2}L^{3/2}T^{-1}$.

Dosage in Bone *

3.5. *The special case of bone.*

(a). *General considerations.* The case of the boundary region between bone and soft tissue requires special consideration because of its importance in clinical radiotherapy at conventional peak potentials of 200 to 300 kv. The presence of bone in a treatment zone alters the distribution of dose in two ways. The macroscopic distribution is altered, with a reduction in the dose reaching soft tissues beyond the bone and, on a microscopical scale, the dose to soft tissues immediately adjacent to or enclosed within bone is raised by the action of secondary electrons arising from photoelectric absorption in the calcium and phosphorus of the bone mineral.

The anatomy of the situation is indicated in figures 3.3 to 3.5, and dose distributions are illustrated purely qualitatively in figures 3.6 and 3.7.

With 200 to 300 kv radiations, the dose enhancement produced by secondary electrons from compact bone will occur only within distances of the order of 100 microns or less from the bone surface. The tissues of interest in such a case are, for example, periosteal tissues at points B C D and corresponding endosteal sites B′ C′ D′ in figure 3.4. Within the bone, the tissues of interest such as E in figure 3.5a will be the soft tissue components of the Haversian system: the osteocytes, the connective tissues lining the walls of the Haversian canals, and the contained blood vessels (see fig. 3.5b). In all these cases the absorbed dose depends upon the linear dimensions of the tissue inclusion relative to the range of the emitted photoelectrons.

The general features of the dose distribution in the region of compact bone are illustrated in figures 3.6 and 3.7. In figure 3.6 the exposure dose is seen to fall below that measured in a homogeneous medium and in figure 3.7 the absorbed dose is shown to rise within the bone itself. The condition at the interfaces between bone and soft tissue are not shown accurately to scale and the linear extent of the interface region is enlarged for the sake of clarity.

(b). *Absorbed dose near an interface and within bone.* The absorbed dose in the compact bone itself is given by

$$D_{\text{mineral bone}} = fR \text{ rad} \qquad (3.1)$$

(see eq 8.6) where R is the exposure dose at the site in question and f is the factor for compact bone for the photon energy (table 8.1) or averaged for the appropriate photon spectrum (table 8.2 and fig. 8.7). These figures indicate the values of the coefficient f averaged for some primary and scattered photon spectra within a water phantom and also the values of f measured directly by the method of equivalent gases (Allisy, 1958). This dose to compact bone, however, is not of direct clinical interest because dose levels sufficient to cause biological effects in the living soft tissue components of bone are unlikely to affect the purely mineral parts.

In very small soft tissue inclusions the absorbed dose will approach that given by eq (3.1) because it will be produced mainly by secondary electrons generated in the compact bone. These electrons, however, lose energy at different rates in soft tissue and in bone, and eq (3.1) must be modified to include the ratio of the mass stopping powers for bone and tissue:

$$D_{\text{soft tissue in bone}} = \frac{fR}{(S_m)_{\text{bone}}/(S_m)_{\text{tissue}}} = FR \text{ rad} \qquad (3.2)$$

where

$$F = \frac{f}{(S_m)_{\text{bone}}/(S_m)_{\text{tissue}}}$$

and

$$(S_m)_{\text{bone}}/(S_m)_{\text{tissue}}$$

is the mass stopping power ratio (bone/tissue) for the average secondary electron energy. This approximation is justified because the $(\hat{s}_m)^{\text{bone}}_{\text{tissue}}$ is a slowly varying function of electron energy. The value of $(\hat{s}_m)^{\text{bone}}_{\text{tissue}}$ varies from 0.914 at an electron energy of 10 kev to 0.931 at 200 kev (calculated from table 8.3). Equation (3.2) holds only for soft tissue inclusions which are small compared with the secondary electron range; this means that for most of the energy range where the photoelectric effect in bone is important the tissue dimensions cannot be much more than 1μ (Spiers, 1951; see also fig. 3.10).

In the transition region near compact bone the absorbed dose varies with distance from the interface; approximate calculations for a plane slab of bone and other simplifying assumptions give values of the absorbed dose as illustrated in figure 3.8 (Spiers, 1951). At the interface itself, the value of F is approximately the mean of the values for muscle and bone derived from tables 8.1 and 8.2; at distances from the interface greater than the photoelectron ranges, the value of the factor falls to that for muscle. In figure 3.9 the distances are given at which the increased dose due to the proximity of the calcified layer is reduced to 10 percent of the value of the soft tissue dose.

If the issue included within the compact bone has dimensions greater than about 1 μ, the dose received will be less than that given by eq (3.2) and will vary from a maximum value at the edge of the cavity which contains the tissue to a minimum at the center. Variations in dose across

* Reprinted from National Bureau of Standards Handbook 78, Report of the International Commission on Radiological Units and Measurements (ICRU) 1959. (The numbers refer to paragraphs in the original report.)

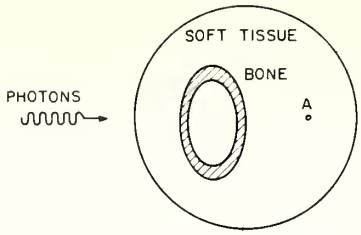

FIGURE 3.3. *Tissues of interest—point in soft tissue shielded by bone.*

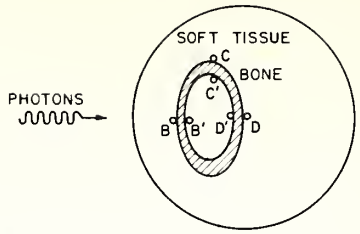

FIGURE 3.4. *Tissues of interest—points adjacent to bone.*

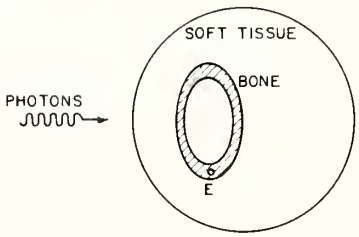

FIGURE 3.5a. *Tissues of interest—points within bone.*

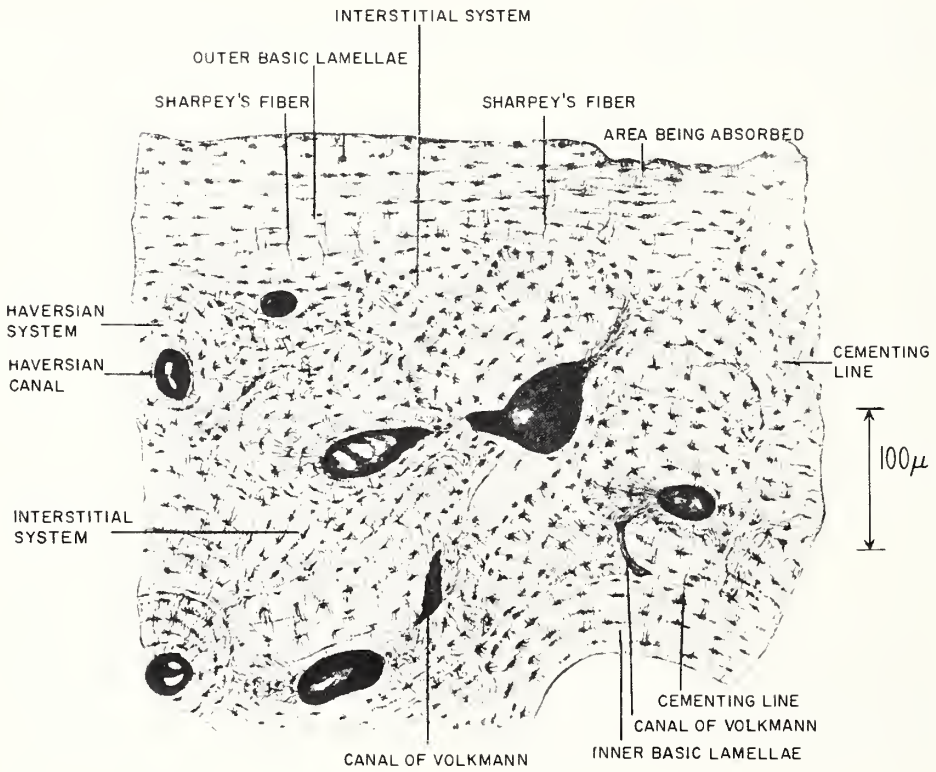

FIGURE 3.5b. *Photomicrograph of a ground portion of human metacarpal bone after staining with fuchsin and mounting in Canadian balsam.*

(After Schaffer in, Maximow and Bloom: Textbook of histology, 5th ed., W. B. Saunders Co., Philadelphia, Pa.)

tissue-filled cavities of different sizes for three photon energies are shown in figure 3.10. The 5-μ cavity may be taken as typical of one containing an osteocyte. The 50-μ and 100-μ cavities correspond to the dimensions of many Haversian canals. These calculations assume that the bone surrounding the tissue-filled cavities provides equilibrium numbers of photoelectrons; in the case of 200- to 250-kvp X-rays, electronic equilibrium is nearly complete if the thickness of bone is 100 μ.

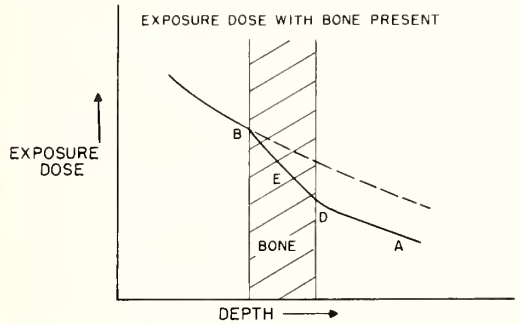

FIGURE 3.6. *Exposure dose with bone present (schematic).*

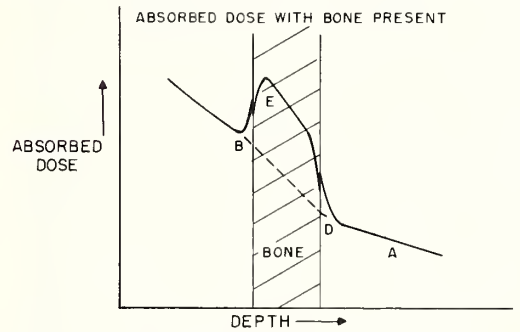

FIGURE 3.7. *Absorbed dose with bone present (schematic).*

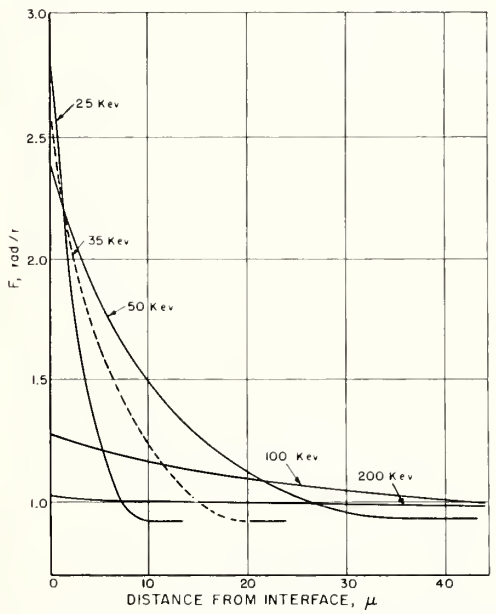

FIGURE 3.8. *Absorbed dose factor, F, in rad/r near bone.*

In the Haversian systems two classes of tissue need to be distinguished: (1) Fixed tissues, and (2) blood cells in transport through the blood vessels. The latter group of cells, however, is only temporarily exposed to the higher dose rate in the canals and, because these cells mix with other blood cells, the absorbed dose they receive will not be significantly raised by the photoelectric effect in bone. The fixed tissues, on the other hand, will receive extra absorbed dose depending on their size and location. They include blood vessels having two types of wall: (1) Endothelial cells only—bone capillaries; and (2) mixed tissues—veins, arterioles, and arteries. It can be seen from figure 3.8 that the endothelium of the vessels in the second category will not

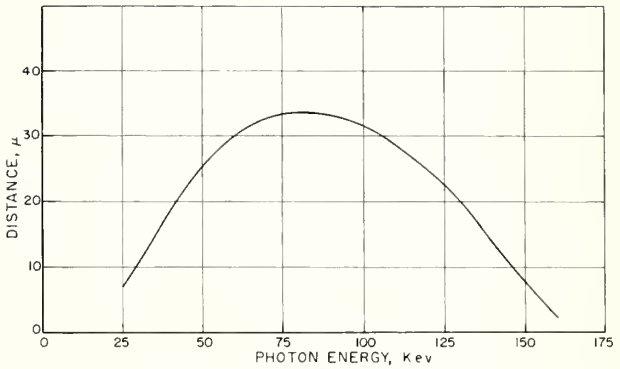

FIGURE 3.9. *Distance at which excess dose falls to 10 percent of soft-tissue dose.*

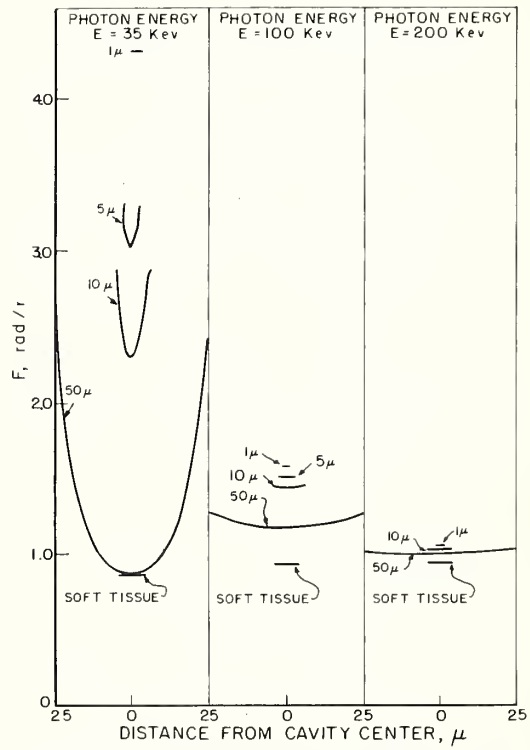

FIGURE 3.10. *Absorbed dose in tissue-filled cavities.*

receive more than about 10 percent excess absorbed dose at any quality of radiation, provided they lie centrally in canals or spaces greater than some 100-μ diameter. The endothelial cells of the bone capillaries in a 50-μ canal, however, could receive a minimum dose of about 140 rads for every 100 rads received in soft tissue remote from bone, when irradiated by 200- to 250-kvp X-rays. At 80 kvp (mean photon energy 35 kev), the absorbed dose to the endothelial cells could be as high as 280 rads (fig. 3.10).

It is clearly difficult to specify exactly the absorbed dose to the soft tissues in bone because of their variability in size and location. Nevertheless, a useful indication of the order of dose can be given, as in table 3.1, by calculating the absorbed dose for three types of tissue: (1) Average absorbed dose to an osteocyte of 5-μ diameter; (2) the mean of the average absorbed doses to two sizes of Haversian canal, one 10-μ and the other 50-μ in diameter; and (3) the average absorbed dose to tissues assumed to be 10-μ thick lining a Haversian canal of 50-μ diameter.

The osteocyte represents the tissue component which in all probability receives the highest absorbed dose, the mean absorbed dose in (2) is intended as an approximate measure of the average throughout the soft tissue inclusions in bone,

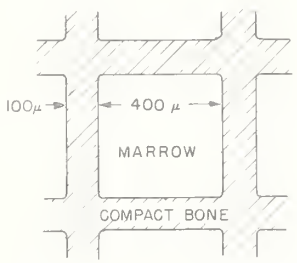

FIGURE 3.11. *Model and dimensions of lamellae and marrow spaces in spongy bone.*

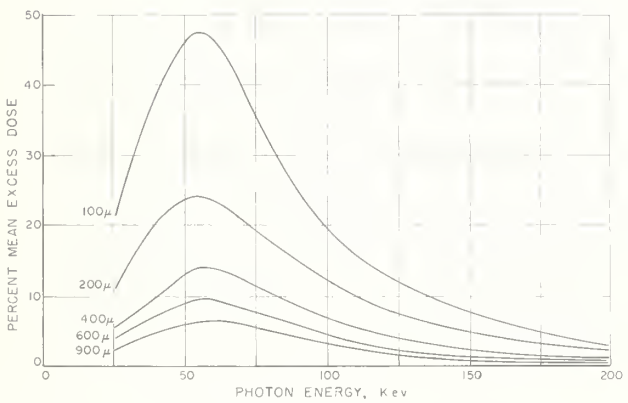

FIGURE 3.12. *Dose excess over soft-tissue dose averaged over marrow cavities.*

excluding the marrow, (Spiers 1951), and the dose to the tissues lining a Haversian canal is given as a possibly significant clinical point.

(c). *Absorbed dose in bone marrow.* The spaces in trabecular bone which contain red marrow are, on the average, considerably larger than the small tissue-filled spaces in the Haversian systems with the result that the mean marrow dose is probably not greatly raised by the photo-electrons from the mineral bone. Engström et al., (1958) give distributions of the sizes of bone lamellae in human vertebrae and ribs as ranging from 60 to 100 μ with the dimensions of the marrow spaces ranging from 250 to 950 μ in vertebrae and 100 to 700 μ in ribs. Robertson and Godwin (1954), studying vertebrae and ribs, give a range of 20 to 250 μ for bone lamellae and 200 to 700 μ for the marrow spaces. Representative mean dimensions might be therefore 100 μ for bone lamellae and 400 μ for the marrow interspaces. Regarding trabecular bone as a formalized lattice as in figure 3.11, the marrow cells filling the spaces will receive a higher marrow dose depending on the size of the marrow space. The mean excess marrow dose, for a range of marrow spaces, has been calculated to vary with photon energy approximately as in figure 3.12. For 400-μ interspaces, the mean excess marrow doses are respectively 5, 13, 7, and 1 percent of the soft tissue dose at photon energies of 25, 50, 100, and 200 kev. Except at the highest energy, where the photoelectric effect is in any case small, the bone trabeculae will be mostly of sufficient thickness to provide electron equilibrium.

(d). *Further considerations and limitations of present analysis.* Tumors of bone have the same atomic composition as soft tissues and where macroscopic tumors have replaced calcified bone the absorbed dose will be the same as for soft tissue remote from bone. Where small extensions of tumor invade the bone structure, or islands of tumor develop in bone by metastatic spread, the absorbed dose will be governed by the same considerations of tissue size and electron range as in the calculations already given.

The evaluation of absorbed dose in bone has been made on the basis of geometrical models and simplifying assumptions such as single photon energies and isotropic emission of secondary electrons. More elaborate calculations have been made in special cases, using complete photon spectra and energy distributions of secondary electrons (Woodward and Spiers, 1953). Further refinements are desirable particularly in relation to the degradation of the primary photon energy spectrum with depth in tissue. Although the change of quality with depth makes little difference to the rad/roentgen conversion factors in soft tissue, in bone the effect can be considerable. For example, the factor $f=1.36$ for "average soft tissue" in bone (table 3.1) irradiated with X-rays of proton energy 100 kev without scatter (as for a finger), would change to 1.67 if a 10×10-cm beam were degraded by backscatter at 2 cm (e.g., a rib) and would increase to 1.87 if the same

primary beam were directed at a bone 10 cm deep. (Data from Greening and Wilson, 1951.) Other data on energy degradation of the spectrum are now becoming available (see section 8.9(a)).

Finally, consideration of the case of high-energy radiation is required. At high-photon energies the pair-production process results in greater absorbed dose in compact bone than in soft tissues. The magnitude of the effect can be estimated, for example, for a photon energy of 10 Mev at which the energy absorption per gram in compact bone is about 15 percent greater than in soft tissue and the positron and electron ranges are some 2 to 3 cm in soft tissue. For equilibrium the bone would have to be more than 1-cm thick and the tissues adjacent to it would then receive an additional absorbed dose of the order of 15 percent. In many cases the bone would be insufficiently thick to give this equilibrium dose, as bony structures represent only a part of the thickness of the anatomical bone; in such cases an average bony composition and density would have to be considered and the additional absorbed dose to adjacent tissues would be in fact less than 15 percent.

(e) *Shielding effect* (at distances greater than the range of secondary electrons from bone).

When bone is present in soft tissue between the radiation source and some position A, the dose rate at A is lower than if the bone is replaced by soft tissue. It is convenient to consider this lowering in terms of the relative reduction, $(D_1-D_2)/D_1$, where D_2 is the dose rate when the bone is in place and D_1 is the dose rate when the bone is replaced by soft tissue.

The dose rate at position A results from the radiation (*a*) that is transmitted directly to it from the source plus radiation, (*b*) which reaches A after one or more scatterings. For D_2 the latter radiation may or may not have passed through the bone while the former has passed through the bone.

The contribution to the relative reduction for radiation (*a*) depends on the relative attenuation of the radiation in the bone and in the same thickness of soft tissue. The relative attenuation of the radiation depends on the density of the bone compared to soft tissue and for low and very, very high (a few Mev) energy photons upon the chemical composition of the bone compared to soft tissue.

The density of bone varies. Spiers (1946) indicates a density of 1.85 for compact bone, Failla (1921), and Jacobson and Knauer (1956) found 1.15 for spongy bone. Haas and Sandberg (1957) found in 12 autopsy specimens of skull caps densities ranging from 1.53 to 1.92, 8 of them being between 1.6 and 1.69 and the average for the entire group being approximately 1.65. They found the same value of 1.65 in one specimen of the mandibular angle. Gest (1959) found 1.65 for a mandible, 1.45 for cervical-vertebrae (Atlas and 7th) and 1.3 for the wing of the iliac bone. Ellis and Jones (1957) found in adult post mortem vertebral bodies a density of 1.1 for the bulk of this bone, while the more anterior portion had an average density of about 1.4.

The problem is thus very complex and when high precision is required the attenuation coefficient must be calculated in each case considering: the characteristic of the bone, the energy of the primary X-rays and the characteristics of the scattered radiations which vary with area of the beam and the depth of the bone layer in the tissue. Furthermore, the change of the scatter contribution due to interposed bone has to be considered. A few examples of this type of calculation have appeared in the literature (Spiers, 1946; Meredith, 1951).

The actual magnitude of the shielding effect progressively decreases as the beam employed becomes harder. For high-energy X and γ radiation the variations of the mass attenuation coefficient with the composition of the bone and its depth in tissue are small and often can be neglected, and any change in the scatter contribution may also be neglected. For all field sizes the effective mass attenuation coefficient of bone is, to a first approximation with high-energy photons, equal to the effective mass attenuation coefficient of water or of soft tissue. Knowledge of the thickness of the bone and of its density makes possible the evaluation of the relative reduction of the dose beyond the bone.

At 200 kvp the relative reduction of the dose due to the interposition of bone varies between 10 percent and 30 percent for bones of thickness varying between 1 and 3 cm (table 3.2).

TABLE 3.1.—*Mean doses to soft tissue components of bone*

Photon energy	Mean dose factor F=rad/r		
	Osteocyte 5-μ diam.	"Average soft tissue" in bone	10-μ lining of 50-μ Haversian Canal
kev			
25	2.80 (3.49)[a]	1.73 (1.94)[a]	1.50
35	3.12 (3.75)[a]	2.05 (2.42)[a]	1.76
50	3.25 (3.65)[a]	2.27 (2.71)[a]	1.89
75	2.40 (2.55)[a]	1.85 (2.10)[a]	1.60
100	1.52 (1.57)[a]	1.36 (1.47)[a]	1.26
200	1.05 (1.06)[a]	1.03$_5$ (1.05)[a]	1.02

[a] Kononenko (1957) has derived analytical expressions for the dose in cavities bounded by plane, spherical and cylindrical surfaces in a medium containing α-emitters. The solution for the case of the plane boundary is the same as that in which the numerical data in this section are based, including the approximate solution for cylindrical geometry to the photo-electric component of the X-ray dose. No solution was given which would enable the dose to the lining of the Haversian canal to the derived, but approximate calculation suggests that the differences between the "plane" and "cylindrical" geometries would be rather less than those in column 3.

TABLE 3.2.—*Dose reduction due to bone shielding*

Reference	Bone density	Bone thickness	Depth of bone	Tube potential	HVL	Field area	Relative dose reduction
	g/cm^3	*cm*	*cm*	*kvp*	*mm Cu*	*cm^2*	*%*
Haas and Sandberg	1.65	1	0	400	2.12	48	11.4
Jones	1.65	0.9	0	-------	2	100	10
Spiers	1.85	3	6	200	1.5	100	27
Institüt Gustave-Roussy (Dutreix)	1.7	4	0	220	2	100	33

For a given quality of radiation the relative reduction of the dose varies slightly with the geometry as follows:

1. Beam area—The larger the field area, the smaller is the relative reduction. This is due to the fact that in any tissue the effective attenuation coefficient decreases when the field area increases. Furthermore, in the case of bone the relative scatter contribution originating in the bone layer increases with the size of the field. For instance, at 200 kvp, HVL 1.2 mm of Cu, and for a bone layer of 4 cm of thickness and 1.7 density, the relative reduction at a depth of 15 cm beyond the bone is 42 percent for a 3×3 cm field, 40 percent for a 5×5 cm field and, for a HVL of 2.0 mm of Cu, the relative reduction is, for the same-fields, 37 percent, 35 percent.

As this variation is so small, it may often be neglected (see A. Dutreix et al., 1959).

2. Bone depth—there is no variation with the depth of the bone in the tissue at 200 kv, but a small variation occurs at lower energies.

3. Distance between bone and position A—The way the relative reduction of the dose varies with the distance between the bone layer and position A is a function of the area of the field and the HVL, sometimes decreasing, sometimes increasing with the distance. For X-ray beams of HVL equal to or less than about 0.5 mm Cu, the relative reduction decreases with increasing distance between the bone layer and the point considered, the rate of decrease being augmented as the primary radiation becomes softer. For a 2.0 mm Cu HVL X-ray beam the opposite effect is observed (see fig. 3.13).

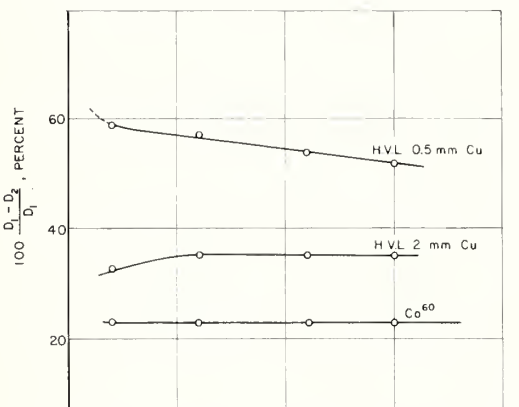

FIGURE 3.13. *Relative dose reduction due to a 4 cm thick bone at different distances beyond bone for various qualities of radiation.*

The processes responsible for these effects are very complex. These are:

(a) A filtration effect which changes the X-ray quality and increases the relative depth dose. This effect is small and negligible at 200 kvp. It is greater at lower energies.

(b) An increase of the forward scatter contribution originating in the bone layer due to the substitution of bone for water and the consequent increased density of electrons to act as scattering centers.

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Appendix III

Estimation of Integral Absorbed Dose*

Though integral absorbed dose may not be regarded as of great importance in clinical radiotherapy, there are occasions when a knowledge of it may be of considerable value. Therefore it is useful to have a simple and generally accepted method for its determination. Of the methods described in the radiological literature (Happey, 1940; Mayneord, 1940; Grimmett, 1942; Meredith and Neary, 1944; Mayneord and Clarkson, 1944; Boag, 1945), those based on calculation seem to be of the most general and the most straightforward application. Furthermore, much of the information required for the calculations may be tabulated as standard constants.

As the report has outlined, various integral dose values may be considered but the most informative appear to be Σ_B, the integral absorbed dose within the confines of the beam, and Σ_R, the integral absorbed dose to the rest of the patient.

Meredith and Neary (1944) give formula for the calculation of integral absorbed dose (per 100 r "in air") under a variety of conditions and one of these formulae can be used for the determination of both Σ_B and Σ_R. For a medium irradiated by a field of surface radius "a", the formula enables one to calculate the integral absorbed dose within a coaxial, confocal cone of surface radius "ka". If $k=1$, the formula gives Σ_B directly. The evaluation of $\Sigma_R(=\Sigma_K-\Sigma_B)$ is more difficult because the variation in size and shape of the body makes it difficult to decide on the value of k which should be chosen.

An extreme would be to calculate Σ_K for $k=\infty$ which would give an upper limit of Σ_R though a rather unrealistic value since a patient can seldom be regarded as an infinite phantom even for the smallest field. A better approach would be to formalize the patient as being a cylinder of 20-cm radius coaxial with the beam axis for all trunk treatments and possibly of 10-cm radius for all head treatments. Σ_K could then be calculated out to this distance. The value of k would, of course, vary for each field size but this should not introduce any difficulty in calculation or tabulation.

The general formula for $k \geq 1$ is

$$\Sigma_K \text{ to a depth } z' = A\int_0^{z'}\left(\frac{d+z}{d}\right)^2 \{P+M[1-2kI_1(\lambda^* a)K_1(k\lambda^* a)]\}dz. \quad \text{(III.1)}$$

When $k=1$ $\Sigma_{K=1}$ is the same as Σ_B

$$\Sigma_B = A\int_0^{z'}\left(\frac{d+z}{d}\right)^2 \{P+M[1-2I_1(\lambda^* a)K_1(\lambda^* a)]\}dz. \quad \text{(III.2)}$$

If $k=\infty$, i.e., we consider an infinite phantom,

$$\Sigma_\infty = A\int_0^{z'}\left(\frac{d+z}{d}\right)^2 (P+M)dz. \quad \text{(III.3)}$$

In these formula the symbols used stand for the following:

A = Field surface area in sq cm
a = Field surface radius in cm
d = Focus skin distance in cm
z = Depth in cm
z' = Depth of phantom
P = Primary percentage depth dose at depth z
M and λ = Meredith-Neary constants for the radiation conditions being used

$$\lambda^* = \lambda\frac{d+z}{d}$$

$I_1(\chi)$ and $K_1(\chi)$ are respectively the modified Bessel functions of the first and second kinds of order one.

It will be noted that each of the above formulas may be expressed in the form

$$\Sigma_K = A\Phi \quad \text{(III.4)}$$

where Φ is the integral of eq (III.1).

Thus Φ may be tabulated for different sets of conditions and the evaluation of the integral absorbed dose for any treatment reduces to the multiplication of the surface area by the value of Φ appropriately selected.

An example of such a table is given below (table III.1), together with an example of its use. In this connection, it must be stressed that the quoted factors allow the calculation of the integral absorbed dose per 100 r delivered "in air." To calculate the values for 100 r "with backscatter," the above calculated values of I_K must be multiplied by $(100/100+S)$, where S is the percentage surface backscatter for the field in question.

TABLE III.1.—*Typical table of* Φ

2.0 mm Cu HVL. 50 cm F.S.D.

z'	$k=1$							$k=\infty$
	Field area in sq cm							All areas
	20	50	100	150	200	300	400	
10	680	800	890	940	980	1030	1070	1300
15	810	990	1120	1200	1250	1330	1390	1800
20	890	1090	1260	1350	1420	1530	1610	2200

*Taken from National Bureau of Standards Handbook 78, Report of the International Commission on Radiological Units and Measurements (ICRU) 1959.

Example. To find the values of Σ_B and Σ_∞ to 15 cm deep for a 10-cm diameter (80 cm^2) field delivering 4000 r with backscatter.

(a) *To evaluate Σ_B and Σ_∞ per 100 r "in air."* From the above table, by interpolation where necessary

$$\text{for } k=1 \qquad \Phi=1068$$
$$\text{and for } k=\infty \qquad \Phi=1800$$

$$\text{Hence } \Sigma_B=80\times1068=\underline{85{,}440} \qquad \Sigma_\infty=80\times1800=\underline{144{,}000}$$

(b) *To allow for backscatter.*
Now the percentage backscatter for this field is approximately 25.

$\therefore \Sigma_B$ and Σ_∞ per 100 r with backscatter will be:

$$\Sigma_B=85{,}440\times\frac{100}{100+25}=\underline{68{,}400} \qquad \Sigma_\infty=144{,}000\times\frac{100}{100+25}=\underline{115{,}000}$$

For 4000 r with backscatter

$$\Sigma_B=2.74 \text{ megagram rads}$$

and $\Sigma_\infty=4.6$ megagram rads.

Hence an estimate of $\Sigma_R=\Sigma_\infty-\Sigma_B=1.86$ *Mg rads.*

The formulas presented can only be applied, of course, where values of M and λ exist; i.e., to those radiation conditions to which the Meredith-Neary formula has been found to fit. For some published data this basic formula does not provide good agreement and is inappropriate to the calculation of central depth dose data and isodose curves. The accuracy, however, required for integral absorbed dose calculations is considerably less than for percentage depth dose data, and therefore constants which give only a moderate agreement between experimental and calculated data could be used satisfactorily in this work.

A test for the accuracy of the method being proposed is not easy to apply since there is comparatively little comparable information in the literature. However, Mayneord (1940), using the equivalent wavelength, calculated that the incident flux per 1 r in air and per sq cm of field surface when an infinite phantom is irradiated by a 10-cm diameter field at 1.5 mm Cu HVL would be 3200 ergs. The Meredith-Neary calculation of the total energy absorbed for such circumstances gives an answer of 3290 ergs. This agreement to within 3 percent must be regarded as most satisfactory, especially when one remembers that Mayneord's computational method for similar circumstances only accounts for about 50 percent of the incident energy.

Though the formula specifically deal with the integral doses for treatments with circular fields, they may easily be applied to rectangular fields by the use in the formula of the radius and area of the "equivalent circles," as evaluated by Jones (1949) and by Day (1950).

In conclusion, therefore, it may be claimed that the method which is most suited for general acceptance as the method for integral absorbed dose determinations is that of Meredith and Neary, because it is probably the most flexible yet described. Further, the necessary constants can readily be tabulated and with them the computations are extremely simple.

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Appendix IV

Techniques for Measuring Radioactivity in Samples and Living Subjects*

4.1. Introduction

This chapter deals with preferred methods of measurement of radionuclides for clinical and biological purposes. Most of the methods for which data are given are well established and are of satisfactory accuracy, particularly for sample counting. However, in the important field of *in-vivo* measurement the accuracy which must be accepted is lower and it is more difficult to recommend standardized procedures. Three aspects of *in-vivo* measurement are discussed: estimation of the radioactivity in the body as a whole, estimation of the radioactivity in an organ (such as the thyroid gland) and determination of the distribution of radioactivity within the body (scanning). These subjects are not treated in detail but wherever possible recommendations or guidance are attempted.

Useful sources of information on these matters include the recent "Manual of Radioactivity Procedures", NCRP–NBS Handbook 80 (1961), and several important symposia reports published by the International Atomic Energy Agency, Vienna. These and other general references cited in the text should be consulted by the reader for more detailed discussions of these subjects and for extensive bibliographies.

4.2. Sensitivity

In selecting the instruments most suitable for a particular measurement, many factors such as convenience and reliability enter into the choice, in addition to sensitivity. Various statistical methods of analyzing data obtained with counting systems are available and may be applied in considering their relative merits. However, stability in the counting system is essential if such criteria are to be valid.

One valuable quantity is the *minimum combined time*, Loevinger and Berman (1951). The total time which must be spent counting the sample and also the background in order to attain a specified precision will be a minimum if

$$\frac{T_{S+B}}{T_B}=\sqrt{\frac{S+B}{B}}$$

Then, the minimum combined time,

$$T(=T_{S+B}+T_B)$$

is given by

$$\sqrt{T}=\frac{1}{e}\left(\frac{\sqrt{S+B}+\sqrt{B}}{S}\right)$$

where S is the sample counting rate corrected for background, B is the background counting rate, and e is the quotient of the standard deviation of S by S. It will be noted that the ratio of minimum combined times with two instruments will be a function of S and B, so that the better of two instruments for measuring weak samples may not be the better with strong samples. When S is small compared with B

$$\sqrt{T}=\frac{2\sqrt{B}}{eS}$$

and hence, since S is proportional to the counter sensitivity, f (the quotient of sample counting rate by activity), it follows that with very small values of S/B the shortest combined counting time is obtained with an instrument for which $\sqrt{B}/f$ is smallest.

The minimum combined time is somewhat complicated, and it is often useful to consider two simpler quantities: the *background equivalent activity*, and the *minimum detectable activity*.

The background equivalent activity, in the particular geometry used, is *that activity of a radionuclide which produces a reading of the instrument equal to its background reading.* (NCRP–NBS Handbook 80, section 2.5.4.)

$$\text{Background equivalent activity}=B/f$$

Minimum detectable activity is defined as *that activity of a radionuclide concerned which, in a given counting time, increases the reading of the instrument by an amount equal to three times the standard deviation of the background recorded in that time.* (NCRP–NBS Handbook 80, section 2.5.3.)

Background and background equivalent activity for typical detectors in four classes of measurement are given later in table 4.1.

The minimum detectable activity is most easily computed for a counter in which the only fluctuations are those due to the random character of radioactive events. For that case

$$\text{Minimum detectable activity}=\frac{3}{f}\sqrt{\frac{B}{t}}=\frac{3B}{f\sqrt{N}}$$

where t is the time spent in counting the background and $N(=Bt)$ is the number of counts accumulated in determining B. It will be seen that, as deduced previously, the best counter at very low activities is that with the smallest value of $\frac{\sqrt{B}}{f}$.

The maximum measurable activities vary greatly with the apparatus, and are less clearly defined. For Geiger-Müller and scintillation counters the

*Taken from Radioactivity, International Commission on Radiological Units and Measurements, Report 10c, National Bureau of Standards Handbook 86. (Numbers refer to paragraphs in the original report.)

maximum measurable activity is in the range of 10^3 to 10^4 times the background equivalent activity, the upper limit being set by the acceptable dead-time correction. For large ionization chambers the maximum measurable activity may be about 10^8 times the background equivalent activity, the upper limit being set by the acceptable recombination loss. In practice the working range is usually limited by the associated current-measuring device (NCRP–NBS Handbook 80, footnote to table 7.1–1).

4.3. Preferred Methods of Measurement of Radionuclides

Techniques of measuring radioactivity in hospitals and biological laboratories are under continuous improvement with regard to sensitivity, accuracy, and convenience. The latter is particularly important in studies involving large numbers of samples. Improvements are generally accompanied by increasing complexity of the apparatus, the subtleties of which can result in erroneous measurements if their operation is not fully understood. Table 4.1 and the accompanying notes present detailed information on the more important measurement techniques which may assist users to choose the methods most suitable for their purposes.

In table 4.1 the measurements are grouped into the following four classes:

I. High-level sources. Measurement of samples before administration to patients or animals.

II. *In-vivo* activity. Measurement of activity in the patient or animal (quantity or distribution).

III. Low-level samples. Measurement of samples or specimens from the patient or animal.

IV. Monitoring. Ancillary measurements concerned with laboratory monitoring and waste disposal.

About half the data quoted in table 4.1 has been taken from NCRP–NBS Handbook 80, tables 7.1–1, 7.1–2, 7.2–1, 7.6–1, and appendix B. The remainder has been assembled from various sources and then checked against current values in at least one radioisotope laboratory (Department of Radiology, Stanford Medical School). The data are intended to represent values which are reasonable at the present time, but not necessarily the best that can be accomplished with each technique.

(b) The visual field at the working distance (20–30 cm) should be preferably 12 cm and certainly not greater than 15 cm in diameter (fig. 4.1).

(c) Collimation and shielding should be such that, if a point source of I^{131} is moved at the working distance, the counting rate falls to 50 percent or less as the distance from the axis increases by 20 percent, and to 5 percent or less as this distance increases by a further 20 percent. With a further movement of the source away from the axis, the counting rate should fall rapidly and remain below 1 percent of the maximum value (fig. 4.1). Figure 4.2 shows the upper limits of counting rate outside the field as a fraction of that in the field.

(d) The contribution of scattered radiation from the patient to the total counting rate should be reduced to a minimum. This could be done either by placing in front of the detector a lead filter of 1.5-mm thickness or by setting the electrical threshold at a suitable level, such as 280 kev. Electrical thresholds should be carefully checked.

(e) The volume of the solution to be used as the "standard" should be similar to that of thyroid glands frequently measured, e.g., a polyethylene bottle approximately 30 mm in diameter and filled with 30 ml of solution will be satisfactory. This bottle should be placed in a cylindrical neck phantom of 15 cm in diameter and of 15 cm in height, made preferably of lucite or persper. The recommended position within the phantom of the

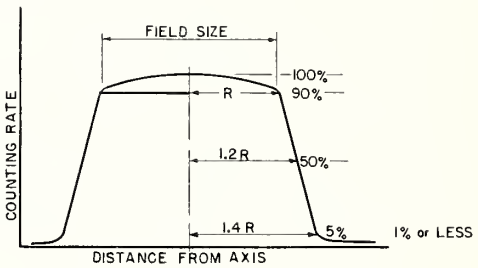

FIGURE 4.1. *Relotive counting rate of detector vs distance between the axis of the collimator and a point source at the working distance (see fig. 4.2) for an adequately designed collimator.*

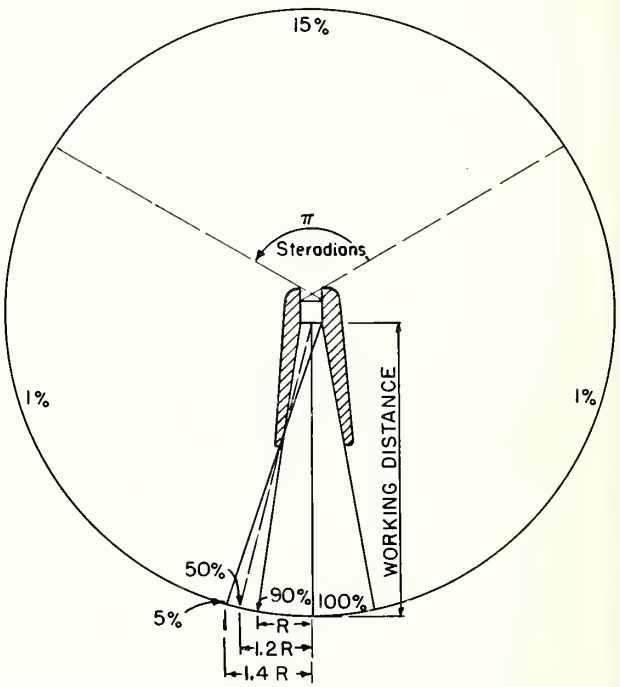

FIGURE 4.2. *Schematic drawing of crystal and collimator and of upper limits of counting rates outside the field of vision.*

TABLE 4.1.—*Data on preferred methods of measurement and radionuclides for clinical and biological use*

Class	Purpose	Footnote	Detector	Background equivalent activity (μc)						Background B (cpm)
				Beta-gamma		Pure beta			Bremsstrahlung	
				I^{131}	Co^{60}	H^3	C^{14}	ρ^{32}	ρ^{32}	
I. Higblevel sources.	Calibration of bulk shipments & therapy quantities.	1	Calibrated well-type ion chamber.	1.	0.2	—	—	0.04	100	(10^{-14} amp)
	Checking of aliquot for administration.	2	Simple, fixed geometry G-M counter.	4.	1.	—	—	—	200	40
		3	Simple, fixed geometry scintillation counter.	0.1	0.2	—	—	—	10	200
	Dose-rate measurement on sealed sources.	4	Fixed geometry ion chamber	100	100	—	—	—	*	(leakage)
		5	Extrapolation ion chamber	—	—	—	100	100	—	(noise)
II. *In vivo* Activity.	Uptake in tissue	6	Fixed geometry scintillation counter.	0.2	0.3	—	—	—	20	900
	Whole-body counting	7	One or more liquid or solid scintillation counters.	*	*	—	—	—	—	*
	In vivo Distribution	8	Scintillation counter and mechanical scan.	0.04	—	—	—	—	—	60
		9	2 scintillation counters in coincidence.	—	—	—	—	—	—	*
	Relative surface activity.	10	Small, hand-beld scintillation counter.	0.01	0.02	—	—	—	0.6	100
		11	Small, end-window G-M counter.	*	*	—	—	*	—	10
	Relative activity within tissue.	12	Needle-probe G-M counter	10	*	—	—	$3\times10^{-4}\frac{\mu c}{ml}$	—	12
		13	Needle-probe scintillation counter.	*	*	—	—	*	—	*
III. Lowlevel samples.	Unprocessed excreta and other large samples.	14	Scintillation counter	0.07	0.05	—	—	—	30	1400
		15	Ring of large G-M tubes	0.1	0.1	—	—	—	20	1000
		16	Well-type ion chamber	3	1	—	—	—	300	(10^{-13} amp)
	Blood and other small samples.	17	Well-type γ-scintillation counter.	2×10^{-4}	2×10^{-4}	—	—	—	9×10^{-3}	200
		18	Well-type β-scintillation counter.	—	—	—	—	4×10^{-5}	—	10
		19	Liquid scintillator	—	—	1×10^{-3}	4×10^{-5}	—	—	100, 40
	Dry Samples	20	Thin-wall liquid G-M counter	5×10^{-4}	8×10^{-4}	—	—	4×10^{-5}	—	10
		21	End-window G-M counter	5×10^{-5}	5×10^{-5}		8×10^{-4}	5×10^{-5}	—	20
		22	Gas-flow G-M counter	1×10^{-5}	1×10^{-5}	1×10^{-4}	1×10^{-4}	1×10^{-5}	—	10
	Chromatographic and autoradiographic samples.	23	End-window G-M counter	5×10^{-5}	5×10^{-5}	—	8×10^{-4}	5×10^{-5}	—	20
		24	Gas-flow G-M	1×10^{-5}	1×10^{-5}	*	1×10^{-4}	1×10^{-5}	—	10
IV. Monitoring.	Routine laboratory monitoring.	25	End-window G-M counter	1×10^{-4}	1×10^{-4}	—	2×10^{-4}	1×10^{-4}	—	40
		26	Scintillation counter	0.02	0.03	—	—	—	2	700
		27	Simple, fixed-geometry G-M counter.	4	1	—			200	40
	Disposal	28	Simple, fixed-geometry scintillation counter.	0.1	.02	—	—	—	10	200
		29	Calibrated ion chamber	3	1	—	—	—	300	(10^{-13} amp)

—not applicable.
*no suitable data available.

hole for the bottle is illustrated in figure 4.3.

For the detailed version of these and other recommendations of the IAEA group, as well as for some comments intended to show the reasons behind some of the points made, the original text of the recommendations should be consulted.

Notes on Table 4.1

1. The data are for one design of well-type ionization chamber which has been developed by the National Physical Laboratory, Teddington, England, and is commercially available (Dale, Perry, Pulfer, 1961, Dale, 1961). These chambers are manufactured to close tolerances and are all of equal sensitivity to within 1 percent for γ rays and 3 percent for β rays. Calibration factors are available for a number of radionuclides. Activities up to several hundred millicuries can be measured, and with conventional electrometers the minimum detectable activity corresponds to approximately 10^{-14} ampere, hence the minimum detectable activity is comparable with the background equivalent activity. Another well-type chamber for this purpose is commercially available in France (Engelman, 1960).

2 and 3. Any reproducible system is satisfactory, provided source-detector distance is not so short that positioning errors are important. If previous measurements are made by the supplier, the accuracy required is not high as the purpose is to check for order-of-magnitude errors and sample identity. (A well-type ion chamber (notes 1 and 16) can be conveniently used for this purpose also). The quoted sensitivities refer to distances of 20 cm.

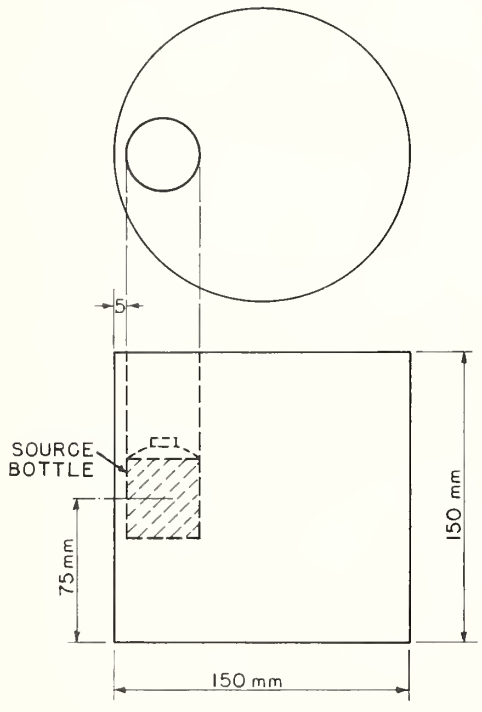

FIGURE 4.3. *Schematic views of the neck phantom and source bottle.*

3. The data apply to a scintillation crystal, 2.5x2.5 cm. used with discriminator at 50 kev, distance 20 cm.

4. The data apply to an R-meter ionization chamber which gives exposure rate directly, but the chamber-source distance should not usually be less than 5 times the largest ion-chamber dimension. The well-type ion chambers (notes 1 and 16) can be very conveniently used for relative measurement of gamma-emitting sources of the same radionuclides, or of different radionuclides if the chambers are effectively "air equivalent" or are properly calibrated.

5. The extrapolation chamber is an absolute instrument, in the sense that exposure rate is determined directly in terms of geometrical and current measurements only. It is a specialized instrument, and its use requires special care and experience.

6. The sensitivities refer to a 5x5-cm crystal, at 25 cm with discriminator at 50 kev. A larger-diameter crystal reduces the minimum detectable activity, but does not appreciably change the background equivalent activity. A pulse-height analyser allows rejection of scattered and background radiation, and improves the sensitivity, no matter how this is specified. Either a scaler or a precision count-rate meter can be used for straight uptake measurements. For measurement of rapidly changing activity (e.g., cardiac output) a count-rate meter with a linear or logarithmic recorder may be used. It is necessary that both instruments have a fast time response, in order not to distort the response curve.

7. A variety of scintillation systems is used. See section 4.

8. The sensitivity for I^{131} refers to a 5x5-cm crystal with a 19-hole focusing collimator, and a 100-kev window centered at 364 kev; however, window widths for scanning with I^{131} may be as small as 40 kev.

9. Measurement of the coincidences of the photons of β^+ annihilation requires 2 scintillation crystals and suitable coincidence circuity. See section 4.5.

10. The sensitivity refers to a 1.25x1.25-cm crystal at 2 cm without discrimination. The relative γ-ray activity of two superficial regions (e.g., palpable thyroid nodules) is best examined with a small scintillation crystal with minimal shielding placed directly in contact with the areas. A simple power supply and either scaler or count-rate meter is adequate.

11. The relative β-particle surface activity of two regions (e.g., the anterior surfaces of the eyes) is best measured with a small end-window G-M counter. No shielding is necessary. A simple scaler or count-rate meter is adequate. The absorption of β particles is so great by even thin layers of over-lying tissue that it is difficult to give meaningful figures for the background equivalent activity.

12 and 13. The relative *in-vivo* β-particle activity inside a tissue (e.g., brain) is best measured with a fine probe. Both G-M and scintillation probes are commercially available. A simple

power supply and either scaler or rate meter is adequate.

12. The figure for I^{131} is for a distance of 10 cm, while that for P^{32} is in $\mu c/ml$.

14. The sensitivities refer to a 4.4x5.0-cm crystal, without a pulse-height analyzer, and a one-liter sample placed directly on top of the crystal. With this technique the reproducibility of the geometry, including the sample volume, is very important. A discriminator is useful to eliminate low-energy background counts, but it should not be used to eliminate radiation scattered in the sample or detector. A discriminator setting of about 50 kev is usually satisfactory.

15. This technique has the advantage of being simple, stable, inexpensive, and insensitive to variations in source position. The G-M tubes can be connected in parallel.

16. The sensitivities refer to a well-type ion chamber constructed to accept a sample volume up to 4 liters (1 gal) at a time. While less sensitive than the preceding method, it is simple, stable, and insensitive to source position. It has the advantage of a very wide range of response, being suitable for sources up to several hundred millicuries. With conventional electrometers the minimum detectable activity corresponds to approximately 10^{-14} ampere, hence the M.D.A. $\approx$ B.E.A./10.

17. The sensitivities refer to a 4.4x5.0-cm crystal, containing a well 1.9-cm diameter and 3.8-cm deep, with discriminator set just above phototube noise. A discriminator is adequate for single radionuclide counting. The discriminator should be set well below the photopeak, e.g., 200 kev for I^{131} and Cr^{51}, and 800 kev for Co^{60} and Fe^{59} are reasonable settings. For bremsstrahlung counting, the discriminator should be set just above the phototube noise. For multiple-tracer measurements, a pulse-height analyzer is almost essential.

18. The data apply to a 5-ml sample in a plastic test tube. Due to absorption of the β particles the results are very sensitive to sample volume and container-wall thickness. The advantage of an organic β-sensitive phosphor is its low γ-ray sensitivity and, as a result, its very low background. It is essential to eliminate low-energy background pulses with a discriminator. Since no peaks are found, it is very difficult to calibrate this detector in terms of energy.

19. This is a very sensitive method of counting, expecially useful for the very low-energy β particles of tritium. The method requires elaborate counting apparatus and refrigeration. The liquid phosphor is easily poisoned (quenched) by an incompatible sample, so careful use of control samples, with and without activity, is necessary. The higher background applies to the counting conditions for tritium.

20. The data refer to a counter with glass walls 30 mg/cm^2 thick. This detector is usually a thin, jacketed, glass-wall tube for active liquids. These are stable and inexpensive G-M tubes and are particularly suitable for beta particles of higher energy. They are, however, also very useful for lower energy β-γ emitters such as I^{131} and even for Co^{60} where the γ rays alone are recorded. Thin-wall metal G-M tubes are also available for "wrap-around" sources.

21. The sensitivities refer to a counter with a 1.4 mg/cm^2 window. This detector is simple, stable, inexpensive and is outstanding for those techniques for which it is suitable. Halogen-filled tubes offer indefinite life, and a simple scaler and power supply are adequate. Reproducible geometry, including mass of sample, is essential.

22. The sensitivities refer to a windowless counter. This method is stable and reproducible and has the advantage of high geometrical efficiency, as a result of placing the sample inside the counter. Contamination of the detector is sometimes a problem. Special gases are usually used, but many counters will operate satisfactorily at higher voltages with natural gas (i.e., illuminating gas).

23. Comments on the detector in note 21 apply here. Chromatograms and autoradiograms can be examined manually, using a large detector with an aperture, or a small detector. Automatic paper chromatogram scanning is provided by commercial equipment which moves an active strip past the detector slit synchronously with the chart motion on a linear recorder, the latter receiving the output of a count-rate meter. The linkage between the active strip and the chart must be very reliable, or interpretation will be difficult. It is desirable that the recorder should periodically indicate fixed points on the chromatogram strip.

24. Comments on the detector in note 22 and on the automatic scanner in note 23 apply here. For very low-energy β particles, especially tritium, it is necessary to have the active paper inside the counting gas. A convenient form of automatic gas-flow chromatogram scanner moves the paper through a narrow gap between two opposed counters, thus counting both sides of the paper.

25 and 26. A very simple scaler or count-rate meter is adequate for laboratory monitoring, though it is very desirable that it have an audible signal. It is generally most convenient if this equipment is a portable survey meter. For monitoring laboratory benches and glassware for β-particle contamination, the end-window G-M tube with a window of about 1.5 mg/cm^2 is best for all β-particle emitters except tritium. For monitoring γ rays alone, a G-M counter may be used, but a scintillation crystal is much more sensitive.

25. The data apply to the same detector as in notes 21 and 23, but for laboratory monitoring the detector is assumed to be lightly shielded, so the background is taken to be twice as large.

26. The data apply to the same detector as in note 3, but the discriminator is set just above the phototube noise, the distance to the source is 2 cm and the detector is assumed to be lightly shielded so that the background is taken to be four times as large.

27 to 29. For waste-disposal monitoring, it is only necessary to make an estimate of the activity. For this purpose a relatively insensitive, fixed-geometry detector is suitable.

27. The data apply to the same detector as in note 2.

28. The data apply to the same detector as in note 3 but the discriminator is set just above the phototube noise.

29. The detector is the same as in note 16 and is larger than the detector described in note 1.

4.4. Thyroid I^{131} Uptake Procedure

The measurement of radioactivity within the body of a patient or animal, class II in table 4.1, is much more difficult than the measurement of the discrete samples encountered in classes I, III, and IV. In general lower accuracy must be tolerated and it is more difficult to recommend detailed procedures that might be acceptable to everyone. However, in measurement of thyroid uptake which is a very widely practiced medical procedure, some degree of standardization of procedures would be highly desirable in order that uptake values may be compared from laboratory to laboratory.

In recent years, several groups have considered the various factors which determine the accuracy of an uptake measurement and some have made recommendations concerning procedure (Brucer, 1960; NCRP-NBS Handbook 80, 1961; IAEA, 1962). The most recent recommendations are those of a group convened by the International Atomic Energy Agency which met in Vienna in November 1960 and considered primarily uptake measurements with I^{131} carried out 24 hours after the administration of the dose (IAEA, 1962). The group recommended a method of measurement which would ensure that the results obtained would be comparable with those reached at other institutions using the same method, yet still provide adequate accuracy, even under rather primitive conditions. These recommendations are endorsed because it may be preferable to achieve agreement among laboratories than to strive for the highest accuracy.

The recommendations of the IAEA group, which deal with the physical aspects of the procedure, may be summarized as follows:

(a) In order to minimise the quantity of I^{131} to be administered to the patient, a scintillation counter with a crystal size of not less than 2.5x2.5 cm should be the instrument normally used.

4.5. In-vivo Distribution Studies (Scanning)

4.5.1. Introduction

The determination of distribution of radioactive material *in vivo* by automatic scanning is a rapidly growing field and it is as yet too early to recommend standardized procedures. However, it is possible to consider some aspects of scanning, including some of the features that good scanning equipment should possess.

Scanning is performed with gamma-ray-emitting or bremsstrahlung-producing radionuclides and conventional circuitry, or with positron-emitting radionuclides and coincidence circuitry. These various techniques of scanning are reviewed in NCRP–NBS Handbook 80 (1961), and in IAEA-WHO "Medical Radioisotope Scanning" (1959). The present discussion is limited to automatic gamma-ray area scanning with a moving detector.

4.5.2. Detector

Gamma-ray emitters of low energy are commonly employed in scanning and for these sodium iodide crystals are the detectors of choice. The diameter of the crystal is limited by practical considerations of shielding weight, and the thickness is limited by the consideration that there is little gain in sensitivity beyond about 5-cm thickness (for I^{131}), while the background increases in proportion to thickness. Crystals 5x5 cm are a good compromise, although both larger and smaller crystals have been used successfully.

4.5.3. Collimators and Shielding

The design of the collimator controls the region seen by the detector. Since collimators generally provide improved resolution at the expense of sensitivity, at least three interchangeable collimators covering a range of resolutions and sensitivities will be found useful. For example, Dewey and Sinclair (1961) have shown that, under certain well defined conditions, a collimator having a very high resolution (e.g., a 91-hole focussing collimator) offers the best chance of detecting radiation from a region of less than 0.5 cm in diameter (and high activity is necessary); for regions between 0.5 cm and 1.5 cm in diameter a lower-resolution and higher-sensitivity collimator (e.g., a 19-hole focussing collimator) is best; for regions between 1.5 cm and 3.0 cm in diameter a more modest resolution and higher sensitivity collimator (e.g., a 7-hole focussing collimator) is best; and above 3.0-cm diameter the lowest-resolution and highest-sensitivity collimator (a single-bore collimator) provides the best chance of detection. The properties of a focussing collimator are that it has a sharp resolution and high sensitivity for small sources at the focal distance; therefore the distance between the collimator and the region to be detected is critical. These properties, however, lose their advantage in practice when the source is large. A slit collimator has advantages for profile scanning. Definitions and data on the resolution of collimators are given in NCRP–NBS Handbook 80 (1961), Brownell (1958), Dewey and Sinclair (1961), and Beck (1961).

In collimators of small angle of acceptance it is of the utmost importance that shielding of the remainder of the field should be of ample thickness; otherwise radiation penetrating the shield may destroy the resolving power. Some commercially available collimators are deficient in this respect.

4.5.4. Mechanical Features

A variety of mechanical means of moving the detector relative to the subject have been devised. The most widely used type of area scanner employs a detector which is moved over a region of the subject in a series of equally spaced parallel lines.

A versatile scanner should have the following mechanical characteristics:

(1) variable-scanning area up to about 40x70 cm;

(2) facilities for interchanging collimators;

(3) means of identifying the detector position with respect to the patient;

(4) variable line spacing;

(5) variable speed covering at least the range 0.1 to 1 cm/sec. Control over scan speed and line spacing makes the scanner useful for large areas, where with higher speed and wider spacing it should be possible to complete a scan on a patient in no more than 30 minutes. The line spacing chosen also depends on the resolution of the collimators used. Experience has shown that the speeds indicated cover the most practical range. However, the influence of scan speed on resolution and detectability has not been thoroughly studied. In certain situations, it might be found useful to scan the region of interest in a second plane at right angles.

4.5.5. Electronic Features

The detector is followed by the usual preamplifier and scaler. A rate meter may also be useful for rapid selection of recording characteristics. Binary scalers are preferable to decade because they provide a finer choice of scale factors. Mechanical recording methods generally use scale factors of 4 to 1 and higher due to the limited stylus response speed and the limited response range of the recording method. Scale factors should increase in steps up to a ratio of perhaps 1000 to 1 to allow for all levels of activity up to large therapeutic doses. The use of a discriminator or a single-channel analyser is valuable in excluding scattered radiation and thus improving contrast in the final display.

4.5.6. Recording Systems

The recording method plays an important role in defining the limitations of scanning techniques. In addition to the conventional solenoid-operated punch method of recording, many techniques have been developed for giving improved visibility to small areas of increased or decreased count rate. One method is the "background erase" technique, in which the counts are recorded only if the count rate exceeds a preset level, but great care is needed to avoid being misled by an unsuitable choice of the preset level. A variation of this technique involves the use of a "backgroung subtract" circuit, whereby a uniform pre-set count rate is subtracted from the entire scan pattern. In photoscanning, the counter signal actuates a flashing light to which photographic film is exposed. If conditions are properly chosen so that a modest increase in the original signal produces a great increase in the density on the photographic film, the contrast of the original record is enhanced on the photoscan. Photocopying methods may also be used to erase background and increase contrast on scan recordings without destroying the original recording. One scanner has been developed using a multi-colored ribbon between the solenoid operated punch and the recording paper, the color selection depending on count rate.

All of these methods involve loss of some of the information originally received by the detector. The best use of the equipment is achieved in systems such as that in which each impulse is stored on magnetic tape. This can then be played back with various levels of background suppression or contrast enhancement until optimum conditions are found.

4.6. Whole-Body Counting

4.6.1. Introduction

This procedure has so far found two main applications, (a) the identification and quantitative estimation of radionuclides present in the body and (b) the study of absorption and retention in the body of intentionally administered radionuclides or labelled compounds (Radioactivity in Man, 1961; Whole Body Counting, IAEA, 1961; Spiers, 1957). The choice of the method of measurement and of the equipment depends on the application.

4.6.2. Methods of Measurement

Identification of an unknown radionuclide requires spectral analysis of the radiations emitted from the subject under study. These are best detected by a scintillation material with high spectral resolution, usually a sodium iodide crystal. Spectra so obtained are distorted by self absorption and scattering, within the body tissues, of the gamma rays (or, in the case of beta-ray emitters, of the bremsstrahlung). The extent of this distortion depends on the location of the activity within the body and on the energy of the radiations emitted. The spectrum is also affected by the presence of other radionuclides in the body such as the naturally occurring K^{40} and the fallout component Cs^{137}.

Once the radionuclide is identified, quantitative estimation of the amount present requires calibration of the detecting device against known amounts of the radionuclide in question using a phantom, or of a short-lived radionuclide with similar metabolic and radiation characteristics as that administered to the subject (e.g., K^{42} has been used for calibration of K^{40} measurements and Cs^{132} is likely to be equally useful for calibration in Cs^{137} measurements). The volume and shape of the phantom should resemble that of the subject studied. Some attempts are being made to simulate uneven

distribution of the radionuclide within the body by using compartmentalised phantoms. A further reduction of the effects of uneven distribution can be achieved by placing both the subject and the phantom in such a counting position that sensitivity of the detecting device is similar for nearly all parts of the body. This is attempted either by arranging a number of crystals symmetrically around the body (or even by completely surrounding it by a liquid scintillation material) or by placing the subject in an arc with a single crystal at its center.

Calibration of other radionuclides is complicated by the presence of K^{40} and Cs^{137} in the body. Their contribution to the counting rate depends on their amount, on body build, and on the energy of the radiations emitted by the radionuclides studied in cases where the characteristic spectra overlap. If corrections for the presence of K^{40} and Cs^{137} in the subject are made using calibrations obtained with phantoms, it is necessary to make some rather crude assumptions based on data obtained in a large number of normal subjects counted under identical conditions.

The detection and quantitative estimation of naturally occurring radionuclides in human subjects is particularly important at low levels and has been primarily responsible for the development of whole-body counting techniques. In this type of study, for example the study of natural K^{40} body burdens, shielding is of paramount importance in order to achieve the lowest possible background. Shielding of the detector itself, leaving an aperture for viewing the patient, is not sufficient because the patient's body will cause some net background alteration. This alteration may be a reduction, or in some circumstances, an increase due to scattering with a resultant change in spectral distribution. A shielded room or enclosure constructed of low-activity materials is therefore very desirable.

In studies with intentionally administered radionuclides the situation is somewhat less difficult, since a "background" spectrum for the subject can be obtained before the radionuclide is given. In such studies it is usually desired to follow the retention of the radionuclide in the body and to express the amount retained as a fraction of the initial burden. However, during the first few hours or even days after administration, the pattern of distribution of the radionuclide may undergo considerable change due to repartition between different body compartments and to specific accumulation in certain tissues. Thus, due to variations in geometry and self absorption, the spectrum changes, and the total counting rate may even increase despite an actual loss of radioactivity from the body. Unless very careful attention is paid to positioning the subject so as to minimize these effects, data for body retention may be meaningless.

4.6.3. Equipment Requirements

The amount of radioactivity involved in the measurement of body burdens is usually quite small and the sensitivity of the detecting device should therefore be high. Scintillation detectors are the instrument of choice. If high spectral resolution is also required to identify one or several unknown radionuclides, sodium iodide crystals are to be preferred. Some workers favor a single large crystal because resolution is better and calibration problems are perhaps less troublesome; others prefer multi-crystal systems, which are advantageous in permitting simultaneous partial body analysis (International Directory, IAEA, 1962).

Most shielding rooms or enclosures are constructed of lead-lined iron; other shielding materials such as talc, chalk, and other minerals with low K^{40}, thorium and radium content are also in use. Care should be taken to avoid the use of steel which, during its production, has been contaminated with Co^{60}. The counting room should be well ventilated with filtered air to maintain a low level of radon daughter products within the enclosed volume. Aged air from storage tanks has occasionally been used to reduce the level of radon daughters or of A^{41} given off by nearby gas-cooled reactors. The room must be kept scrupulously clean and it is helpful to line the surface with a low activity synthetic plastic; subjects to be counted should be showered and scrubbed to remove surface contamination, and provided with monitored clothing. With small body burdens, counting times of 30–60 min. may be required to arrive at reasonable counting statistics; claustrophobia of the subject may be a problem.

The spectrum obtained is recorded by a multichannel analyzer. Channel stability is most important. Fully transistorized instruments are now available which require little attention and can be coupled with various automatic devices, such as tape recorders, computers and data printers, which may greatly ease the task of processing the wealth of information obtained. If it is desired to subtract one spectrum from another automatically, special techniques may be needed to keep energy discrimination very stable; a gain shift of more than one-half per cent might introduce gross errors in the analysis.

If the radionuclide present in the body is known, then good energy resolution is no longer essential and can be sacrificed in favor of sensitivity or simplicity. Several institutions use liquid-scintillation counters made in the form of a pipe which surrounds the body and to which a large number of multiplier phototubes is attached; a distinct advantage of this arrangement is the short counting period. The use of a multichannel analyzer is no longer required.

In metabolic work, knowledge of retention data alone is of little value unless supplemented by other measurements on rates of distribution or organ uptake. This usually requires blood-plasma-assay procedures, and hence the sensitivity of additional apparatus for small sample measurement determines the amount of radionuclide to be administered. Thus the high sensitivity of the devices for whole-body counting described above

may be of little advantage in metabolic work unless sample-counting equipment of adequate sensitivity and stability is provided. Simple whole-body counters can frequently be used, and useful results have already been obtained without an iron room, employing a single-channel analyzer only.

4.6.4. Some Applications

In monitoring radiation workers, whole-body counting is a useful technique for assaying contamination relative to the permissible body burden; likewise the effectiveness of various decontamination procedures can be ascertained with high sensitivity. On various occasions, routine surveys of workers have revealed the presence of minute amounts of such radionuclides as Zn^{65}, Cs^{137}, Pu^{239} and others to which the subjects have unknowingly been exposed under normal and, presumably, safe working conditions. Another application involves the assay of total body Na^{24} induced by accidental exposure of a subject to a neutron beam, although much information on the neutron dose can also be obtained by simply analyzing a plasma sample for its Na^{24} content.

The technique has also been used for estimation of contamination (with radium and its daughter products) of watch-dial painters and, other radium workers, as well as with thorium and its daughter products of patients who had received a colloidal compound of thorium for diagnostic purposes. Such measurements can be usefully combined with analyses of exhaled breath for radon or thoron content respectively. It has recently been found that various groups of dial painters have become contaminated with substantial amounts of Sr^{90}; these cases are at present being carefully studied and attempts are being made to use whole-body counting of bremsstrahlung to assay total body burdens.

Whole-body-counting techniques are growing in use in clinical research work. Here their main advantage lies in effectively replacing the troublesome and invariably incomplete collection of excreta over long periods of time. Na^{22}, Cs^{47}, Fe^{59}, Sr^{85}, I^{131}, I^{131}-labelled proteins and vitamin B_{12} labelled with various radioactive cobalt isotopes have been used to study retention in the body.

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Recommendations [a] of International Commission on Radiological Units and Measurements (ICRU)

ICRU Report Number	Reference [b]
1	Discussion on International Units and Standards for X-ray work Brit. J. Radiol. **23,** 64 (1927)
2	International X-ray Unit of Intensity Brit. J. Radiol. (new series) **1,** 363 (1928)
3	Report of Committee on Standardization of X-ray Measurements Radiology **22,** 289 (1934)
4	Recommendations of the International Committee for Radiological Units Radiology **23,** 580 (1934)
5	Recommendations of the International Committee for Radiological Units Radiology **29,** 634 (1937)
6	Report of International Commission on Radiological Protection and International Commission on Radiological Units National Bureau of Standards Handbook 47, Washington, D.C. (1951)
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8	Report of International Commission on Radiological Units and Measurements (ICRU) 1956 National Bureau of Standards Handbook 62, Washington, D.C. (1957)
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10b	Physical Aspects of Irradiation National Bureau of Standards Handbook 85, Washington, D.C. [c]
10c	Radioactivity National Bureau of Standards Handbook 86, Washington, D.C. [c]
10d	Clinical Dosimetry National Bureau of Standards Handbook 87, Washington, D.C. (1963)
10e	Radiobiological Dosimetry National Bureau of Standards Handbook 88, Washington, D.C. (1963)
10f	Methods of Evaluating Radiological Equipment and Materials National Bureau of Standards Handbook 89, Washington, D.C. [c]

[a] Current recommendations are included.
[b] References given are in English. Many of them were also published in other languages.
[c] In preparation.